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A SURVEY OF ADVANCED TECHNIQUES FOR ACOUSTIC IMAGING.(U)
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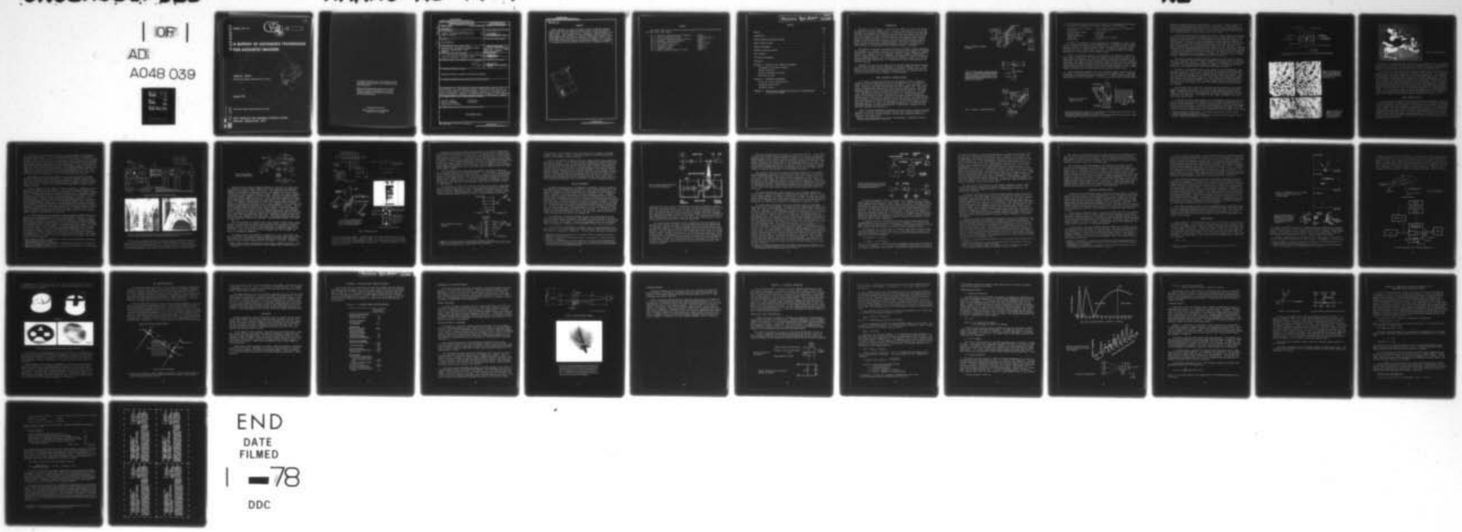

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A SURVEY OF ADVANCED TECHNIQUES FOR ACOUSTIC IMAGING

JAMES M. SMITH
MATERIALS TESTING TECHNOLOGY DIVISION

August 1977

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ARMY MATERIALS AND MECHANICS RESEARCH CENTER
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ABSTRACT

This document is a state-of-the-art survey of advanced acoustic imaging techniques developed for nondestructive testing and medical applications. Several techniques are evaluated: simple mechanical scanning systems, array imaging, acoustic holography, synthetic aperture, Bragg imaging, and ray tracing. Examples of practical imaging systems that employ these techniques are described and compared, and in order to put these systems into proper chronological perspective, an appendix is included that briefly discusses older acoustic imaging techniques.

PREFACE

The author would like to thank the following people for kindly supplying illustrations used in this report:

Dr. G. Knollman, Lockheed Research Laboratory	Figure 2
Dr. C. Quate, Stanford University	Figures 3a, 3b, 3c
Dr. S. D. Ramsey, Stanford Research Institute	Figures 5a, 5b
Dr. M. Maginness, Stanford University	Figure 6
Dr. G. Kino, Stanford University	Figures 7c, 7d
Dr. H. D. Collins, Holosonics	Figures 9, 10
Dr. C. Burckhardt, Roche	Figure 11
Mr. G. Posakony, Battelle	Figure A-2
Mr. H. Berger, National Bureau of Science	Table 1

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CONTENTS

INTRODUCTION

The purpose of this report is to acquaint the reader with some of the recent developments in acoustic imaging systems. Because of the multiplicity of systems that have been proposed and developed, it is not possible to review each and every technique, but hopefully the imaging systems that have been chosen for discussion will give insight into current research efforts in this rapidly expanding field. Most of the emphasis recently has been toward medical imaging systems, primarily because of the research money available, but also because of the huge potential market for medical ultrasonic imaging equipment. Very little research work is being directed toward imaging techniques applied exclusively to nondestructive testing (NDT), but useful fallout from the medical field is inevitable.

This report is primarily a descriptive account of acoustic imaging techniques, and the emphasis is directed toward the critical features that separate one approach from another. Many different techniques are discussed: simple mechanical scanning systems, array imaging, acoustic holography, synthetic aperture, Bragg imaging, and ray tracing techniques. It is hoped that the reader will come to realize that numerous approaches are available for converting ultrasonic signals into a usable image and that each approach must be evaluated in the light of the particular application desired.

Appendix A has also been included that may provide useful background information for those not familiar with ultrasonics or acoustic imaging.[1] Here, a brief survey of early acoustic imaging techniques is presented. In addition, the characteristics of piezoelectric transducers (Appendix B) and the resolution and sensitivity of pulse-echo systems (Appendix C) are discussed.

SIMPLE MECHANICAL SCANNING SYSTEMS

It is useful to begin the discussion by considering a simple, mechanically scanned C-scan system that might be used for NDT applications. The system in Figure 1a consists of focused piezoelectric transducers, a pulse generator, transmitter, receiver, linear gate and peak detector, and a display unit. The transducers are mechanically scanned in a raster fashion and the received signals are amplified, gated in time by the linear gate, and then peak detected. The output of the peak detector drives a recorder or modulates the intensity of a storage scope. Note that the positional information from the scanning transducers is coupled to the display so that a two-dimensional image is formed corresponding to discontinuities in the acoustic impedance at the focal region. The system of Figure 1a operates in through transmission; however, for many applications, a pulse-echo mode is preferred. Here, one transducer serves as both the transmitting and receiving detector. In general, the performance capability is similar for both approaches although the axial resolution may be substantially improved for the pulse-echo approach (see Appendix C).

There are several performance parameters that can be used to evaluate an imaging system. A few of the more important ones are the resolution (range or axial, lateral), sensitivity, transmitted power, dynamic range, operating frequency, frame rate, and imaging mode.

1. BERGER, H. *A Survey of Ultrasonic Image Detection Methods.* Acoustical Holography, A. F. Metherell, H. M. A. El-Sum, and L. Larmore, ed., Plenum Press, New York, v. 1, 1967, p. 27.

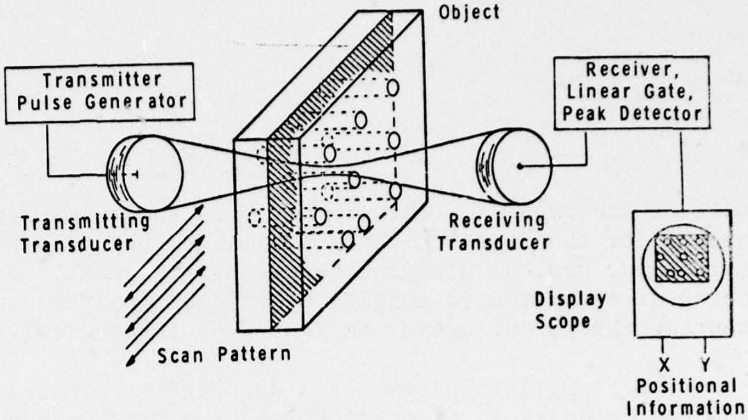
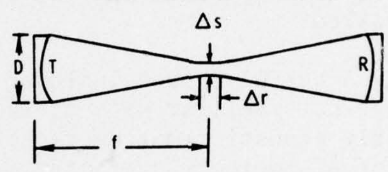


Figure 1a. Diagram of a focused C-scan system.



λ = acoustic wavelength

$$\Delta s \text{ (lateral)} = m\frac{\lambda f}{D}$$

$$\Delta r \text{ (axial)} = m'\lambda\left(\frac{f}{D}\right)^2$$

$$m \simeq m' \simeq 1/2$$

Figure 1b. Spatial resolution of focused transmitter-receiver pair. The equations represent the theoretical limit where beam size is determined entirely by diffraction and aberrations have been ignored. In some applications the resolution will be much worse than predicted by these equations.

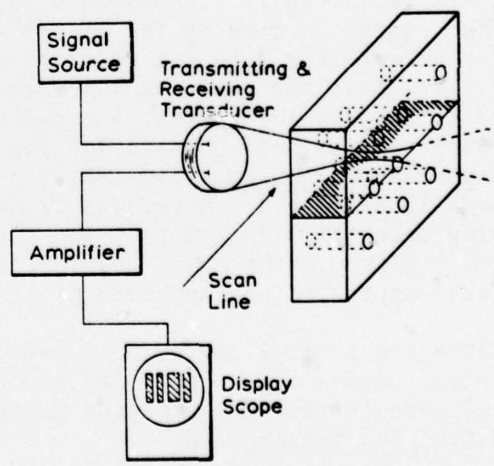


Figure 1c. Diagram of a focused B-scan system.

For the type of approach described in Figure 1a, the performance characteristics of a typical nondestructive testing system might be the following:

Resolution (axial, lateral)	—	a few acoustic wavelengths (see Figure 1b)
Sensitivity	—	10^{-9} - 10^{-10} W/cm^2
Peak transmitted power	—	1-100 W/cm^2
Dynamic range	—	60-70 db
Operating frequency	—	1-20 MHz
Frame rate	—	one frame per 10 minutes
Imaging mode	—	C scan

Of course, the selection of these parameters is determined to a great extent by the particular application of the imaging system. For example, unlike nondestructive testing, a clinical ultrasonic imaging system requires that the transmitted power be kept low, generally below 1 W/cm^2. In addition, high frame rates (30 frames/sec) and a B-scan format (Figure 1c) are often desirable for medical applications. These points will be discussed later.

There is an important property of the focused C-scan approach of Figure 1a that should be emphasized. The high concentration of the acoustic energy at the focal region of the transducers not only improves resolution and sensitivity, but reflections from defects or grain boundaries outside the focal zone are virtually eliminated. This makes it possible to obtain images that are free from the speckle and interference effects normally present when coherent radiation is used. Computations[2] have shown that the image quality from a focused transmitter and receiver compares favorably with techniques employing incoherent acoustic energy.

There are numerous examples of the use of focused C-scan systems for NDT applications. One recent application is the testing of graphite billets conducted by Knollman at the Lockheed Research Laboratory.[3] A typical example of one scanning approach is shown in Figure 2. In order to avoid deleterious imaging effects

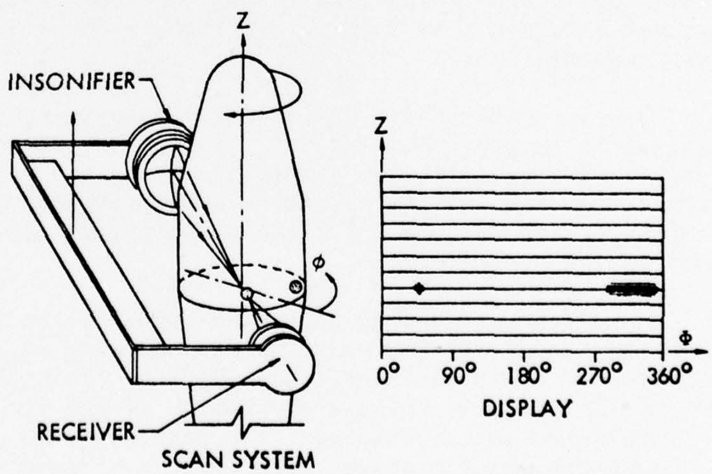


Figure 2. X-Y scanning with a single-element receiver.

2. See Discussion in paper G. Kino, C. DeSilets, J. Fraser, and T. Waugh. *New Acoustic Imaging Systems for Non-Destructive Testing.* 1975 Ultrasonics Symposium Proceedings, IEEE Cat., 75 CHO 994-4SU.
3. KNOLLMAN, G., WEAVER, J., HARTOG, J., and BELLIN, J. *Acoustic Imaging Techniques for Real-Time Nondestructive Testing.* Acoustical Holography. N. Booth, ed., Plenum Press, New York, v. 6, 1975, p. 637.

caused by refraction from the curved surfaces, the billet is either immersed in a fluid with a sound speed nearly equal to the test material, or placed within a solid plane-parallel matching block which has interior surfaces that conform with the cylindrical geometry of the billet. This system has excellent lateral resolution of approximately 1.5 mm and sensitivity of 10^{-10} W/cm^2.

A more novel application of through-transmission C-scan imaging is the development of an acoustic microscope at Stanford University by Lemons and Quate[4] (shown schematically in Figure 3a). This instrument has been operated with acoustic frequencies as high as 1 GHz, with the corresponding lateral resolution approaching one micrometer. Although the operating frequency is much higher than the low megahertz region normally used for NDT and medical diagnostic work, and the objects that are imaged with the microscope are much smaller, the basic approach is nearly identical to the scanning techniques that have been discussed. The object, attached to the mylar film and positioned at the focal point of the transmitter and receiver, is mechanically scanned in the X and Y directions. The acoustic image is formed by intensity modulating the electron beam in a cathode ray tube (CRT), with the beam moving in synchronism with the mechanical motion of the object.

The acoustic images in Figures 3b and 3c compare quite well with the optical ones and, in general, for most biological specimens the image contrast is much better acoustically than optically. There is another quite different type of acoustic microscope which has also produced impressive images. This system is available commercially from Sonoscan and is described in the open literature.[5]

The primary disadvantage of most C-scan systems is that the mechanical scanning process is slow. This is not a problem for the acoustic microscope because the area scanned is so small that a complete scan can be made in one second. However, for most NDT applications, the scanning process takes a few minutes. Scanning schemes that use either an area or a linear array of piezoelectric elements are inherently much faster and appear to be very attractive for NDT applications that require fast inspection rates. Typical examples of such systems will be described later in this report, but first a few words should be said concerning B-scan imaging.

B-scan systems have historically been used much more extensively for medical diagnostic imaging than for NDT. In a simple B-scan system, a cross-sectional image is formed, with the vertical axis of the image representing depth into the sample, and the horizontal axis representing the linear scanning position of the transducer (see Figure 1c). The intensity of the image is related to the amplitude of the acoustic echo.

In a clinical B-scan system (Figure 4), the transducer is coupled to the skin with an aqueous gel and manually scanned with an articulate arm that senses the position of the transducer. In addition, scanning along a curved surface like the abdomen is possible because any motion of the transducer in the plane of the image is reproduced on the display. The returned echo amplitudes are either used to modulate a two-level storage scope (with a threshold level for writing), or are stored

4. LEMONS, R., and QUATE, C. *Acoustic Microscopy – A Tool for Medical and Biological Research.* Acoustical Holography, N. Booth, ed., Plenum Press, New York, v. 6, 1975, p. 305.
5. KESSLER, L., PALERMO, P., and KORPEL, A. *Recent Developments with the Scanning Laser Acoustic Microscope.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 15.

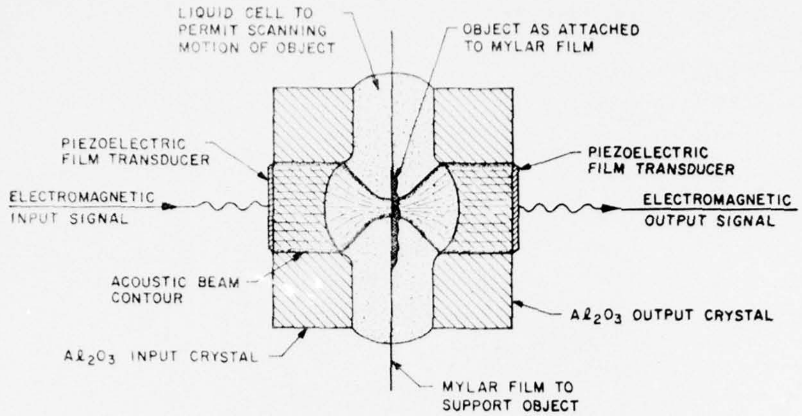


Figure 3a. Schematic diagram of the acoustic system showing the lens configuration.
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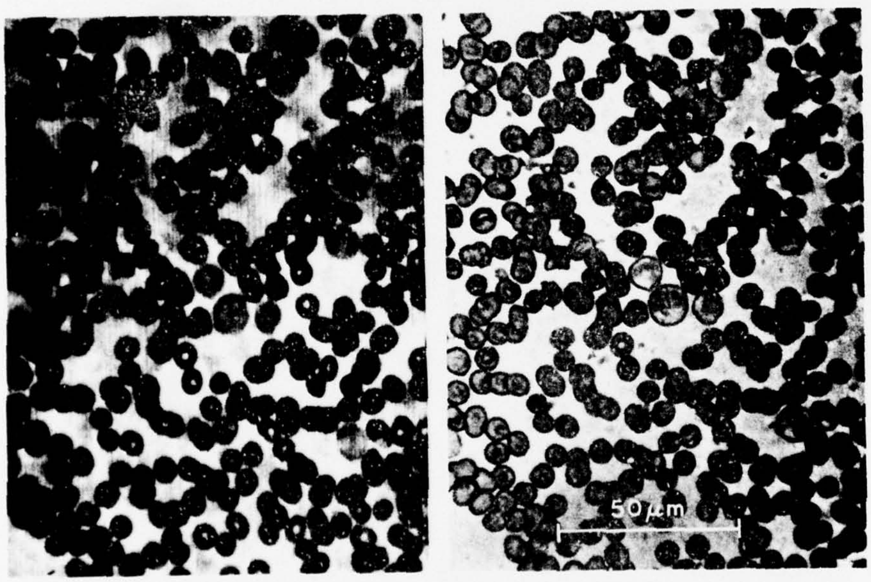

Figure 3b. Comparison of the 900-MHz acoustic image (left) with the optical image (right) of a human bone marrow smear.
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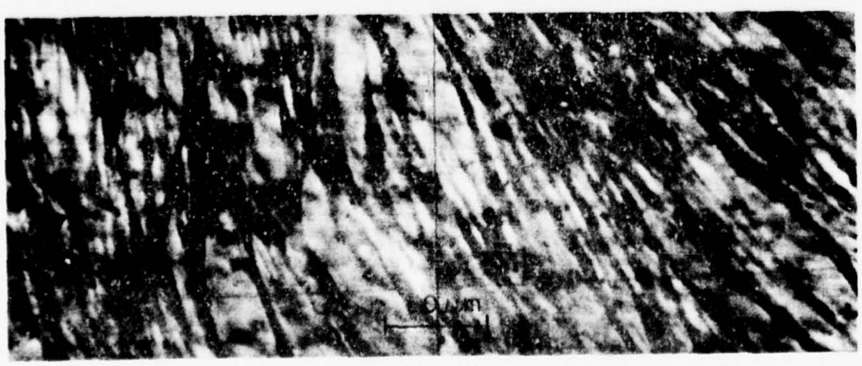

Figure 3c. Acoustic image (900 MHz) of a living culture of normal human diploid lung fibroblasts.
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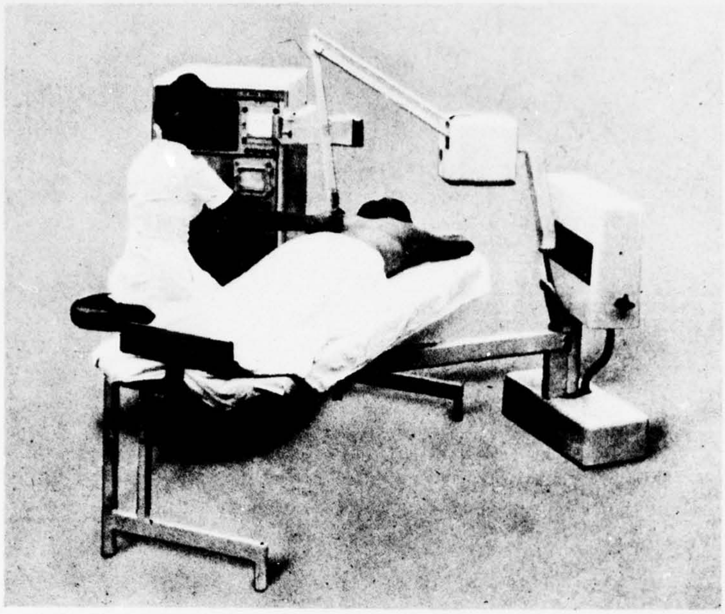


Figure 4. Clinical B-scan system.

in the form of a charge pattern on the face of a scan converter tube. At the completion of the scan, the scan converter, which contains gray scale information, can be read into a TV monitor or into a hard copy unit for a permanent record of the image. Polaroid pictures from the monitor are, of course, also possible.

The clinical B-scan images have a subjective character about them in that the technologist is in a sense "painting" the image onto the display. By rocking the transducer back and forth slightly during the scan, targets can be illuminated by the transducer at different angles resulting in improved image quality. The subjective quality of the image can be remedied by scanning the transducer with a motor or by going to linear array techniques that electronically fire the elements. The transducer array techniques appear to be the most promising, and the development of fast scanning linear array systems for real-time B scanning is actively being pursued by nearly all the major manufacturers of medical imaging equipment. Perhaps in the next year at least ten real-time systems will be available commercially. The development of array systems for NDT applications is proceeding at a much slower pace, primarily because of lack of funding, but it is believed that many of the techniques developed for medical imaging can be applied to NDT.

ARRAY IMAGING SYSTEMS

The primary motivation behind the use of piezoelectric transducer arrays is the desire to reduce the rather long time required to scan an object, particularly in the C-scan mode. By electronically switching between the elements of a linear array, scanning times can be reduced by approximately a factor of one hundred for C-scan systems. For such a system, the linear array must still be either mechanically moved in one direction or the acoustic beam deflected in such a way as to generate a two-dimensional image. An imaging system that employed an area array would eliminate the need for any mechanical movement of the array; but, at the present time, large area arrays with more than a couple hundred elements are well beyond the state of the art.

This section is divided into three parts: unfocused arrays, lens focused arrays, and phased arrays. The unfocused array approach consists of sequentially firing the array elements either individually or in small groups. The earliest commercial system using this technique was perhaps the cardiac imaging system designed by Bom.[6] Recently, a similar B-scan linear array system that is designed primarily for abdominal scanning has become available commercially.[7] This system uses 64 piezoelectric elements, pulsed four at a time in overlapping sequence to produce an image of 61 lines of gray scale information. The lateral and range resolutions are claimed to be 6 and 2.5 millimeters at an operating frequency of 2.25 MHz. The rather poor lateral resolution is typical of this approach and can be improved by focusing techniques, using acoustic lenses or phased electronic focusing.

Two examples of lens focused systems will be given; the Stanford Research Institute's ultrasonic camera developed by P. Green[8] and company, and the acoustically focused area array system developed by Meindl's group[9] at Stanford University. Both of these systems were designed and built for medical imaging, and both systems use acoustic lenses to focus the acoustic energy onto an array of piezoelectric elements.

A block diagram of the SRI camera is shown in Figure 5a. Although the system appears to be complicated, and indeed it is, the critical imaging components are the single-element transmitter, the lenses and prisms, and the 192-element receiving array. The transmitter generates an acoustic waveform that is propagated through the water, scattered by the object, and focused onto the linear array. The operating frequency is 2 MHz, but the instantaneous frequency is swept from 1.5 to 2.5 MHz in order to reduce coherent speckle and interference effects. Perhaps the most intriguing aspect of this system is the deflector assembly which consists of two polystyrene prisms. These prisms, located between the two identical acoustic lens elements, are counter rotated to sweep the ultrasonic field across the receiving array. The resultant C-scan image, generated at 15 frames per second, consists of 400 interlaced lines with 192 elements in each line. The SRI camera, with a lateral resolution of approximately 1.3 mm and a depth of focus of 1 cm, produces acoustic images of high quality (see Figure 5b).

The cardiac imaging system developed by Meindl's group at Stanford University employs a two-dimensional monolithic array of 100 elements (Figure 6). Each element operates independently as its own source and detector. In order to improve resolution, an acoustic lens is used to focus the sound bursts from these elements onto a particular plane through the object. The system uses the latest techniques of integrated circuitry, and is capable of generating real-time images at 30 frames per second in both B- and C-scan modes. The obvious advantage of using an area array is that there is no need for complicated beam deflecting prisms or mechanical scanning in order to produce a two-dimensional C-scan image. However, a C-scan image with Meindl's system consists of only 100 picture elements, compared to approximately 8×10^4 elements in each picture for the SRI camera. Until large two-dimensional arrays become feasible, the image quality of systems using linear arrays will be far superior.

6. BOM, N., LANCEE, C., HONKOOP, J., and HAGENHOLTZ, P. *Ultrasonic Viewer for Cross-Sectional Analyses of Moving Cardiac Structures.* Bio-Med. Eng., v. 6, 1971, p. 500-508.
7. Advanced Diagnostic Research Corporation, Tempe, Arizona.
8. GREEN, P., SCHAEFER, L., JONES, E., and SUAREZ, J. *A New, High-Performance Ultrasonic Camera System.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974.
9. MAGINNESS, M., PLUMMER, J., and MEINDL, J. *An Acoustic Image Sensor Using a Transit-Receive Array.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 619.

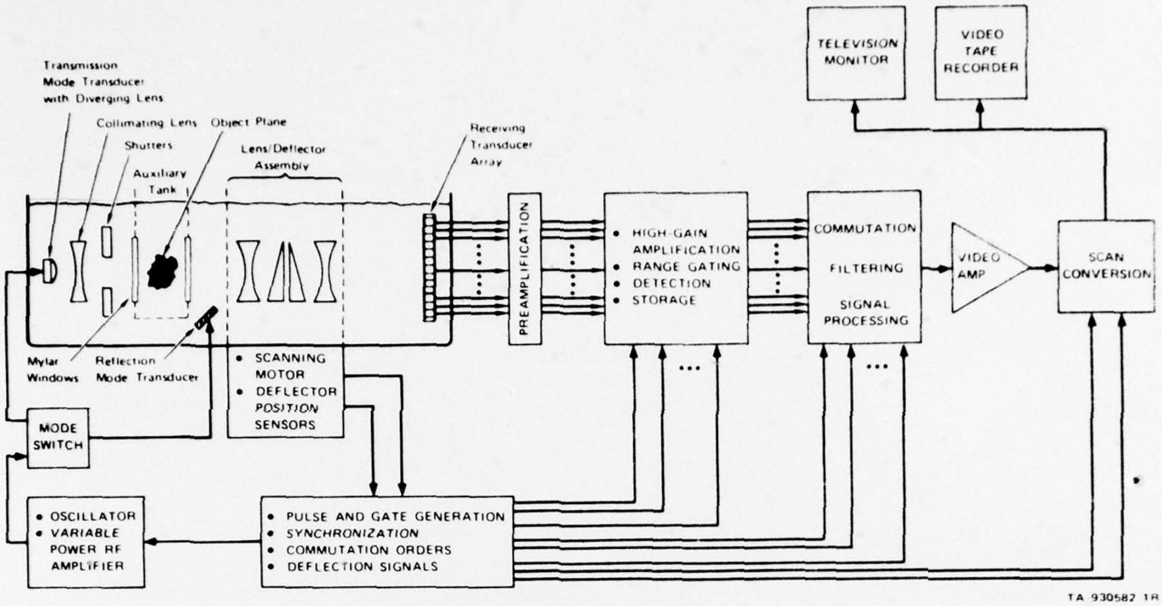


Figure 5a. Block diagram of SRI ultrasonic camera.

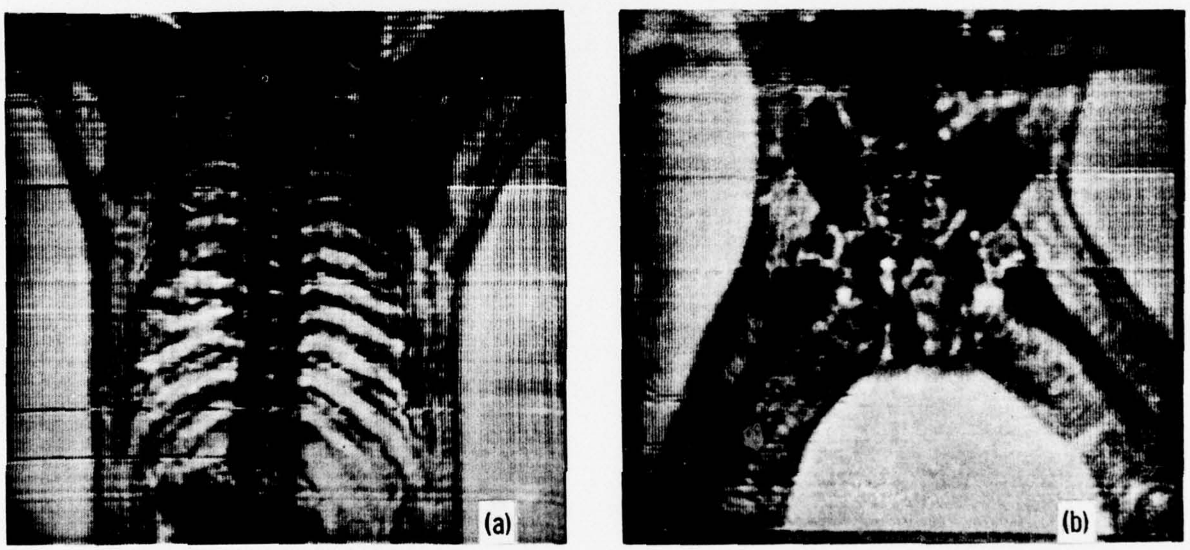


Figure 5b. *Ultrasonic through-transmission images of an eviscerated human infant:* (a) *thoracic cavity and* (b) pelvis. Note the clarity and definition of the skeletal and cartilaginous structures.

The two previous imaging systems use acoustic lenses to focus the acoustic energy onto the piezoelectric array, and the lateral resolution of these systems is indeed superior to an unfocused array. It is also possible to focus an array by electronically delaying or phase shifting the signals from the transducer elements. Two distinct approaches to the problem of electronic focusing will be presented.

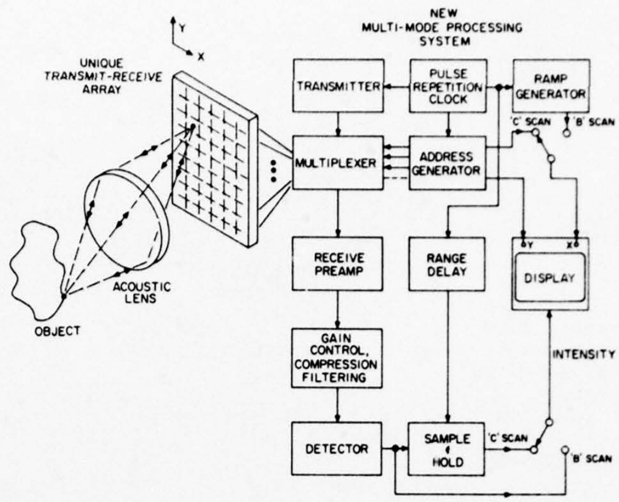


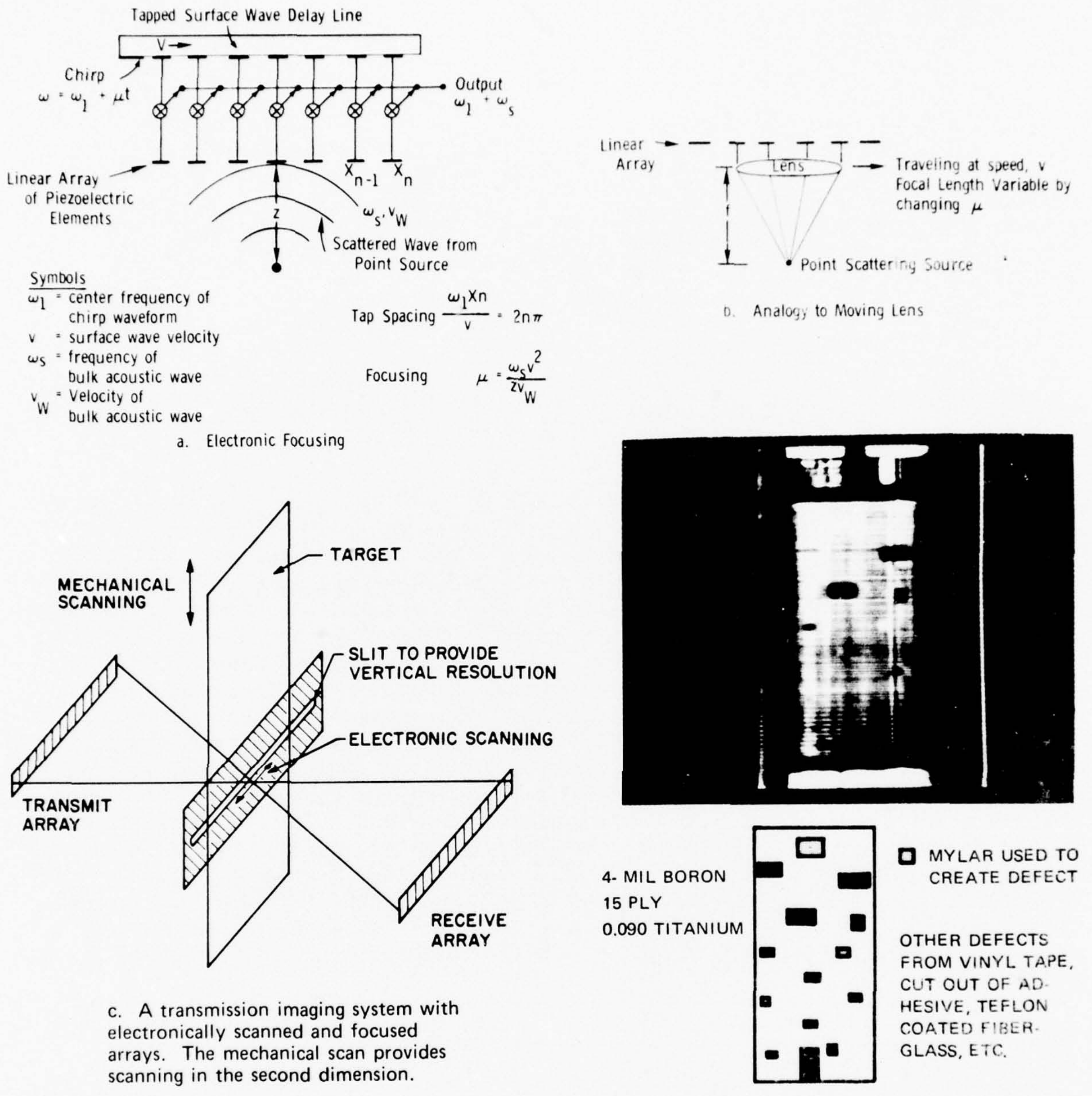
Figure 6. Block diagram of acoustically focused system.

Professor Kino's system at Stanford University[10] consists of a linear array of piezoelectric elements and a tapped surface wave delay which are connected together by the mixers, as shown in Figure 7a. The operation of this system is perhaps best explained by examining how the system operates as a receiver when illuminated by a point source of acoustic energy. Spherical wavefronts, generated by the point source, impinge on the linear array resulting in a quadratic phase variation along the length of the array. If the individual elements in the array were summed together with this phase variation present, the output would be negligible. A technique is needed that will cancel this quadratic phase variation, producing a constant phase along the array so that a summation of the elements will produce a large output voltage when a point scattering source is present. The linear FM modulated waveform (linear chirp) that is applied to the delay line does the required function of cancelling this quadratic phase. This is accomplished because a linear chirp signal has its own time-related quadratic phase variation which is converted to a spatially quadratic phase change by the fact that the waveform travels down the surface wave delay line, interacting through the taps with the piezoelectric elements of the linear array. The required spacing between taps and the relationship between the chirp rate μ and the focal distance z are given in Figure 7a.

The system operates in a fashion that is very similar to what would be accomplished if a moving lens was placed in front of the array, as shown in Figure 7b. However, in Kino's system a line of image information is produced very quickly, determined by the time it takes for the surface wave to travel the length of the delay line. Also, the focal point of this simulated lens can be changed very easily by changing the parameter μ.

The imaging system just described is capable of a B- or C-scan format, and good results have been obtained using two separate 80-element arrays opposite each other (Figure 7c), operating in a through-transmission C-scan mode. Mechanical scanning in the direction perpendicular to the array is, of course, necessary to

10. FRASER, J., HARLICE, J., KINO, G., LEUNG, W., SHAR, H., TODA, K., WAUGH, T., WINSLOW, D., and ZITELLI, L. *An Electronically Focused Two-Dimensional Acoustic Imaging System.* Acoustical Holography, N. Booth, ed., Plenum Press, New York, v. 6, 1975, p. 275.



a. Electronic Focusing

b. Analogy to Moving Lens

c. A transmission imaging system with electronically scanned and focused arrays. The mechanical scan provides scanning in the second dimension.

d. Boron Titanium Laminate Panel

Figure 7. Phased-array system.

form a two-dimensional image. A typical image of a boron titanium laminate panel with induced debonded areas is shown in Figure 7d, and the resolution of the image parallel to the array is approximately 1.3 mm. This technique would appear to have excellent potential for NDT and medical applications, and should eventually be very inexpensive to build.

Another quite different "phased" array approach is the sector scanning technique developed both by Somer[11] and Thurston.[12] Professor Thurston, from Duke University, uses a 16-element linear array that steers and focuses a 2.25-MHz acoustic beam through an angle of 60°. A simplified way of looking at this B-scan system is to imagine an acoustic beam swinging back and forth that originates from the center of the array. The focusing and steering of the beam is accomplished by appropriately delaying the trigger signals of the pulsers that fire the elements in the array, using computer-controlled delay line circuitry. Since a computer controls the scan format, important parameters, such as scan angle and the number of lines per frame, can be readily changed.

The receiving and transmitting modes of operation are shown in Figures 8a and b, and their operation is somewhat different. As a transmitter, the array generates a broad line focus beam; but, when used as a receiver, the array dynamically focuses at each range distance so that the returned echoes arriving at the array elements can be summed in phase. The system is inherently complex because of the need for fast, variable delay lines that are coordinated by a computer. However, the technique works quite well, and commercial units are now available for medical applications.

The resolution for Thurston's system in the plane of the scan is range dependent but typical values are 2 to 6 mm. Sector scanning systems have a great potential for medical applications such as cardiology, neurology, obstetrics, and internal medicine. Their applicability to NDT may be restricted somewhat because of mode conversion problems unless the array is contact-coupled to the object.

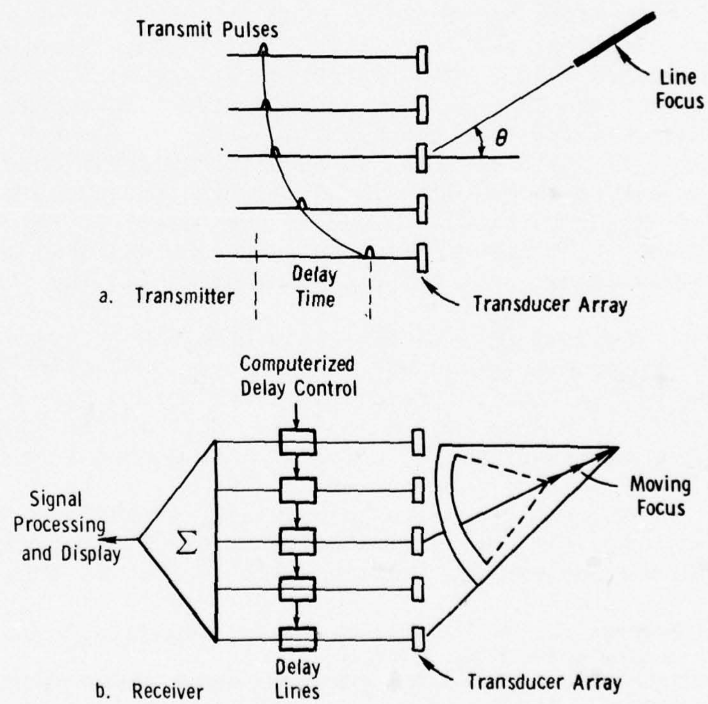


Figure 8. Phased-array transmitter and receiver.

11. SOMER, J. C. *Electronic Sector Scanning for Ultrasonic Diagnosis.* Ultrasonics, v. 6, 1968, p. 163.
12. THURSTON, F. L., and VON RAMM, O. T. *A New Ultrasound Imaging Technique Employing Two-Dimensional Electronic Beam Steering.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 249.

If the object is water immersed, then the scanner will be limited to scanning through a small angle in order to avoid mode conversion from the water to the object in fast acoustic velocity materials.

Many of the problems associated with array imaging systems center around the piezoelectric arrays. It is difficult enough to manufacture single-element transducers with reproducible and well-characterized specifications, but the problem of producing an array of 100 elements that have nearly identical characteristics is even more acute. In many cases, wide band operation of the array is desired, requiring backing material that can be bonded uniformly to the elements without acoustic ringing. Mechanical and electrical cross-talk between elements can also severely limit the performance of an imaging system, and this cross-talk is difficult to avoid because the elements are physically very close to each other. Although these problems have not been completely solved, usable arrays can now be built that perform adequately for a wide variety of acoustic imaging techniques.

ACOUSTIC HOLOGRAPHY

An acoustic imaging system is fundamentally holographic if phase information associated with the returned coherent echoes is used in the imaging process. Because the principles of optical holography were first applied to acoustic applications as recently as the mid-1960's, acoustic holography is still a rather new field. Initially, it was anticipated that three-dimensional images of opaque objects might be produced, much like their optical counterparts; but it was soon realized that the large axial distortion, introduced by the reconstruction process, prevented this. This distortion arises because the optical wavelength of a typical laser that might be used to reconstruct an acoustic hologram is nearly a thousand times smaller than the usual acoustic wavelength employed for NDT or medical imaging. In spite of this, an acoustic hologram does contain information concerning the entire three-dimensional illuminated space, but we must be satisfied in viewing only a particular plane of the object at any one time. In the reconstruction schemes that are usually employed, one can scan the entire depth of an object by simply adjusting the focal point of an imaging lens.

Two commercially available holographic systems will be discussed, both of which use single-element transducers. Holographic systems have been built that use piezoelectric arrays, but these arrays have typically been used in systems that digitally process the holograms. Some of the advantages as well as the disadvantages of holography in comparison to nonholographic systems will be considered.

The liquid surface holographic[13] technique, which was developed nearly ten years ago,[14,15] is available commercially from Holosonics (Model 100,300) for both NDT and medical applications (Figure 9). In this system, an acoustic transducer

13. BRENDEN, B. B. *Real-Time Acoustical Imaging by Means of Liquid-Surface Holography.* Acoustical Holography, G. Wade, ed., Plenum Press, New York, v. 4, 1972, p. 1.
14. BRENDEN, B. B. *A Comparison of Acoustical Holography Methods.* Acoustical Holography, A. F. Metherell, ed., Plenum Press, New York, v. 1, 1967, p. 62.
15. GERICKE, O. R., and GRUBINSKAS, R. C. *Utilization of Liquid Surface Levitation Effect as a Means of Ultrasonic Image Conversion for Materials.* Journal of the Acoustical Society of America, v. 45, no. 4, April 1969, p. 872-880. Note: The technique described in this reference is not true holography, but this work does represent an early application of the liquid surface levitation effect and in that sense is very similar to liquid surface holography.

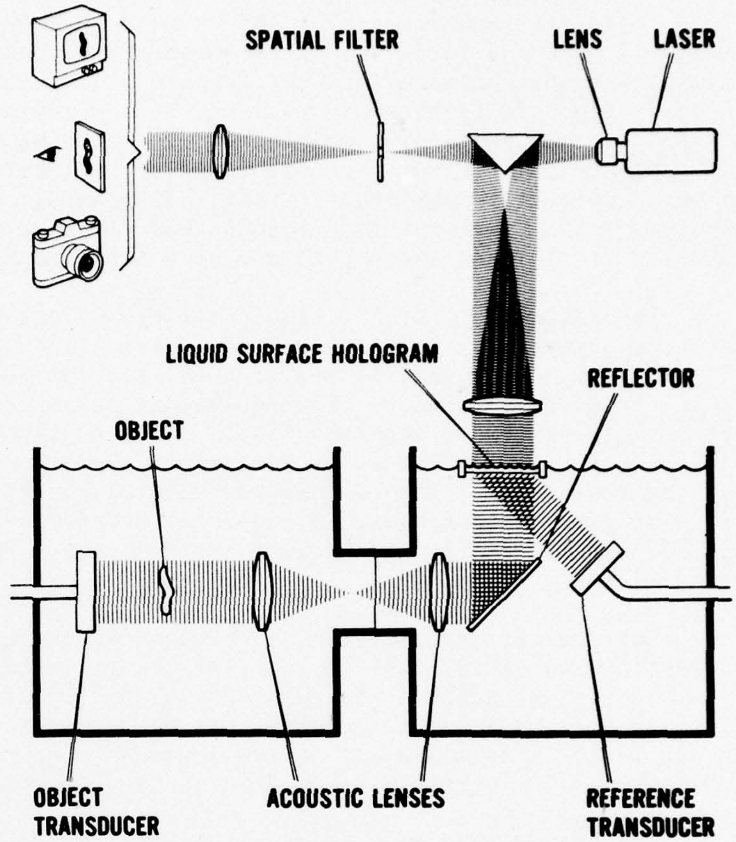


Figure 9. Schematic diagram of Holosonics, Inc. liquid surface imaging systems.

illuminates the object, and the scattered acoustic field is focused by acoustic lenses onto the liquid surface. In principle, the scattered acoustic waves do not have to be focused onto the surface, but better results are obtained if this is done. The focused sound waves are mixed with the acoustic waves from the reference transducer to produce a surface interference pattern, and the DC component of this rippled surface forms the basis of a hologram. The reconstruction of the hologram is achieved by reflecting a laser off the surface, and focusing the first-order diffraction pattern onto a television camera tube.

The liquid surface holographic system is capable of real-time images of an arbitrary plane through the object, and acceptable images for many applications are obtained. However, the images do contain speckle and fringing effects that are not normally present in the nonholographic systems that have been previously described. The term speckle refers to the randomly positional variation in image intensity caused by phase cancellation and reinforcement of the waves hitting the liquid surface. The fringing effects are also an interference phenomenon, which are often caused by diffraction, particularly when an acoustic wave impinges upon a sharp edge. In addition, circular interference fringes are quite noticeable and these occur because the illuminated object is in the near or Fresnel zone of the object transducer, resulting in phase and amplitude variations across the acoustic beam front.

There are techniques that can be used to reduce these effects. If the object is moved back and forth during the inspection process, the eye will integrate the changing fringe pattern, and the fringes will appear to be smoothed out somewhat. Another approach is to rapidly change the operating frequency and superimpose the images so that possible cancellation of the fringes and speckle will occur. Both of these techniques work to some extent, but interference effects are still a serious problem for many applications. In general, all of the holographic imaging systems suffer somewhat from speckle and fringing effects and the images are not usually as clean as those produced by nonholographic approaches.

The sensitivity of the liquid surface holographic approach is not as good as most nonholographic techniques. The interaction between the laser light and the liquid surface hologram is inefficient, and the sensitivity is determined by the noise level of the laser. The equivalent noise temperature of laser light is approximately 10,000 K compared to 300 K for a piezoelectric detector. Therefore, the sensitivity for purely piezoelectric systems should be approximately two orders of magnitude better than liquid surface holography, but in practice the difference is even greater than this. Piezoelectric detectors have a threshold sensitivity of 10^{-11} W/cm^2 but liquid surface systems have a typical sensitivity of 10^{-7} to 10^{-9} W/cm^2.

For medical applications, high sensitivity is very important because the acoustic power level must be kept low enough (usually less than 1 W/cm^2)[16] to prevent biological damage. It becomes very difficult to image with real-time liquid surface holography when such low peak power levels are used. The absolute acoustic power level is normally not critical for NDT applications, and one can usually circumvent the low sensitivity by driving the transducers at high power levels.

The second type of holographic system is sometimes called a simultaneous source-receiver[17] scanning system. The commercial unit manufactured by Holosonics (Model 200) typifies this technique, and this system is shown in block diagrams of Figure 10. Typically, a 3-MHz transducer, focused on the top surface of the object, is scanned under water over a prescribed aperture in a rectangular raster fashion. The area scanned is usually a 10-cm by 10-cm aperture, containing approximately 250 scanning lines. The received acoustic echoes from the internal flaws are time gated and mixed with an electronic reference signal which simulates an off-axis acoustic reference beam. The output of this mixer, which contains phase and amplitude information about the flaw, modulates the intensity of a storage oscilloscope or a glow modulator tube. The oscilloscope requires X and Y positional signals from the transducer, and the hologram is built up during the scanning process on the face of the CRT. In the glow modulator tube system, a light-emitting diode attached to the top of the transducer is modulated, and photographic film is exposed to this light. In either method, a hologram is eventually recorded on transparency film and reconstructed with a laser in much the same fashion as in optical holography. However, there is an important difference in that the image is brought to focus on a ground glass screen and displayed on a TV monitor. This results in a two-dimensional image of the object at a prescribed depth. By adjusting a reconstruction lens, the hologram can be focused at different depths, and in this way the internal structures will appear to come in and out of focus as one scans through the object.

16. WADE, G. *Acoustic Imaging with Holography and Lenses.* IEEE Transactions on Sonics and Ultrasonics, November 1975, p. 385.
17. COLLINS, H. P. *Acoustical Holographic Transverse Wave Scanning Technique for Imaging Flaws in Thick-Walled Pressure Vessels.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 175.

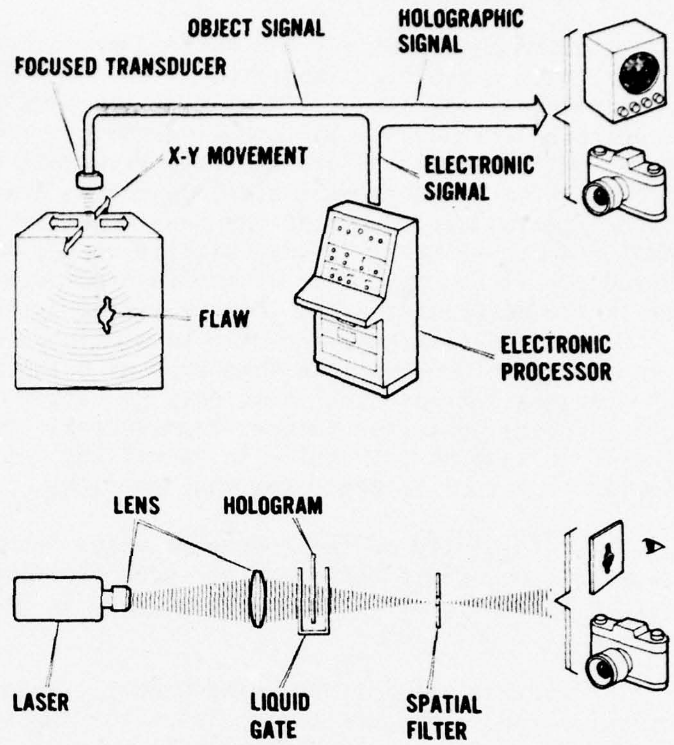


Figure 10. Schematic diagram of Holosonics, Inc. pulse echo acoustical holography systems and holographic reconstruction unit.

The focused transducer, which is typically 1" in diameter with a 4" focal length, plays an important role in this system. This transducer generates spherical wavefronts that illuminate a large volume of the object. This means that the transducer will receive echoes from an internal structure during a large portion of the total scanned area, and the effective aperture of the hologram will be appreciable. The effective aperture increases with flaw depth until eventually it will equal the size of the actual scanned aperture. The size of the effective aperture is important because this is the area of the hologram that contains meaningful phase and amplitude information concerning an internal structure, and the resolution capability of a hologram is directly related to this aperture and not to the size of the actual scanned area.

In addition, the use of a spherical wavefront tends to reduce speckle and fringing effects because internal structures are illuminated at various oblique angles during the scanning process. Interference effects are not nearly as noticeable in this system as they are in the liquid surface holographic system.

The resolution for a system like the Holosonics Model 200 is related to the effective aperture, as previously mentioned. If we assume that the spot size of the focused transducer is negligible, then the lateral and range resolutions are given by the following equations:

$$\Delta s \text{ (lateral)} \simeq \lambda/2 \; z/D_{eff}$$

$$\Delta r \text{ (range)} \simeq \lambda/2 \; (z/D_{eff})^2$$

where z is the distance of the flaw from the holographic plane (the focal point of the focused transducer), λ is the acoustic wavelength in the material, and D_{eff} is the effective aperture of the hologram. Note that these expressions are identical to resolution equations for the coincident focused receiver and transmitter (Figure 1b), with z and D_{eff} substituted for f and D. Therefore, in principle, the two approaches would image a flaw with the same resolution if the size of the focused transducer used in the nonholographic system was equal to the effective aperture of the hologram. Since the effective aperture for the Holosonics unit is typically 3 to 6 inches on a side, depending upon the depth that is being imaged, a large focused transducer of the same dimensions would have to be used in the nonholographic system. We have simplified the analysis somewhat because we have ignored the finite spot size of the focused beam that generates the spherical waveform in the Holosonics unit. For practical systems, this spot size is the limiting factor in the determination of the resolution. However, the important point is that in deep imaging applications this type of holography offers a means of obtaining very fine resolution which could only be achieved in a nonholographic system by employing large-diameter focused transducers. On the other hand, holographic imaging of thin samples (<1" thick in steel) has not been very successful because the effective aperture is small for shallow flaws.

The sensitivity of the source-receiver scanned holographic system is good because piezoelectric detectors are used for transmission and reception. The sensitivity, therefore, compares favorably to nonholographic systems that use piezoelectric detectors.

The optical reconstruction of acoustic holograms could be replaced by digital reconstruction. This would reduce some of the well-known aberrations of the reconstructed image and would perhaps increase the sensitivity. Of course, computer holographic processing has its own set of problems which include such things as computer round-off noise, difficulty in determining correct image plane, and excessive computer time for a complete scan of the object. The time required for reconstruction depends on the computation speed of the computer, the frequency of the acoustic energy, and the number of planar images that are needed. Fast Fourier techniques have greatly reduced processing time and one particular image plane can probably be imaged in a second or so. Another advantage of digital processing is that it is easily adapted to image enhancement schemes which might include spatial filtering, correction of aberrations, smoothing of data information, and pattern recognition.

There are basically two mathematical approaches that have been applied to digital processing of acoustic holograms: lens modeling and backward propagation. The lens modeling approach, used by the Naval Undersea Center[18] for sonar holographic imaging, consists essentially of simulating the focusing function of an acoustic lens by the multiplication of each element in the piezoelectric array by an appropriate quadratic phase factor. The backward propagation technique, employed recently by Fenner of Aerospace[19] and by others,[20] moves the plane of the hologram back to an object plane by adding appropriate phase factors to the plane wave components of the hologram data. This technique requires two fast Fourier transforms rather than the one required for lens modeling, but it is a more versatile approach and usually gives more accurate results.

18. SUTTON, J. L. *An Experimental Focused Acoustic Imaging System.* Acoustical Holography, G. Wade, ed., Plenum Press, New York, v. 4, 1972, p. 351.
19. FENNER, W. R., and STEWARD, G. E. *An Ultrasonic Holographic Imaging System for Medical Applications.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 481.
20. BOYER, A., HIRSCH, T., JORDAN, J., LESEM, L., and VAN ROOY, D. *Reconstruction of Ultrasonic Images by Backward Propagation.* Acoustical Holography, A. F. Metherell, ed., Plenum Press, New York, v. 3, 1970, p. 333.

The quality of the digitally processed images that have been produced so far have not been very impressive, and this is due in large part to the relatively small number of elements in the hologram. With the rapid development of powerful minicomputers and new processing techniques, the full potential of digital processing may be realized in the not too distant future.

Both holographic and nonholographic systems have been discussed, and the choice of the best approach really depends upon the particular application. A source-receiver holographic system would appear to be attractive for deep imaging problems such as nuclear pressure vessel imaging. The resolution would be much better than that achieved with a focused nonholographic system, and the severe spherical aberration associated with trying to focus an acoustic beam through eight to twelve inches of steel would be avoided. A scanned holographic system is also capable of operating in an acoustic interferometry mode which might be useful for detecting minute surface cracks and pits in materials. On the other hand, nonholographic focused imaging offers excellent sensitivity, freedom from speckle and fringing effects, and is ideal for biological imaging or the detection of flaws that are not buried deep in fast acoustic velocity materials.

SYNTHETIC APERTURE SYSTEMS

Synthetic aperture techniques improve resolution by processing the returned echoes in such a way that the effective aperture of the system becomes much larger than the physical aperture of any component in the system. Perhaps the most successful application of synthetic aperture techniques was the processing of radar data in side-looking radar terrain imaging. Here, an aircraft[21] sends out a fan-shaped radar beam that propagates perpendicular to the direction of flight. Although the physical dimension of the antenna is small, the radar data is processed in such a manner that the resolution of the system is equivalent to that obtainable with a large physical antenna, with an aperture or length even larger than the aircraft.

There are many different synthetic aperture techniques that have been applied to acoustic imaging. Even the scanned source-receiver holographic system (Holosonic 200) can be considered to be a synthetic aperture approach because the aperture of the completed hologram is synthesized by piecing together information using a small transducer aperture (diameter of transducer). Other more exotic synthetic aperture schemes have been developed or proposed and many of these approaches use linear arrays in crossed[22] or complicated patterns.

In this report, the use of synthetic aperture techniques to improve the lateral resolution of B-scan images will be discussed. This work, described recently by Burckhardt,[23] was selected as an example because the processing technique is directly analogous to that used for side-looking radar systems. The analysis of synthetic aperture systems can be considered from many points of view: doppler processing, summation, correlation, and holography. Here, the holographic point of view will be stressed.

21. BROWN, W. M., and PORCELLO, L. J. *An Introduction to Synthetic-Aperture Radar.* IEEE Spectrum, September 1969, p. 52.
22. WELLS, W. H. *Acoustical Imaging with Linear Transducer Arrays.* Acoustical Holography, A. F. Metherell and L. Larmore, ed., Plenum Press, New York, v. 2, 1969, p. 87.
23. BURCKHARDT, C. B., GRANDCHAMP, P. A., and HOFFMANN, H. *Methods for Increasing the Lateral Resolution of B-Scan.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1973, p. 391.

In most single B-scan systems, the lateral resolution is poor because one normally uses an unfocused transducer in order to obtain the large depth of field required in the axial direction. The lateral resolution is therefore not much better than the width of the unfocused acoustic beam. The synthetic aperture approach obtains good lateral resolution by generating one-dimensional holograms that have a large effective aperture and therefore good resolution. The acoustic transducer for this system produces a beam with a wide angular divergence, and it is scanned in one direction much like any ordinary B-scan system. Figure 11a shows the scanning scheme and it should be noted that two point targets are to be imaged. The use of a beam with a wide angular spread is critical in order to generate the large desired effective aperture. This requirement is the opposite of that normally encountered in nonholographic systems where a narrow beam width is desired.

In the synthetic aperture B-scan system, the acoustic transducer transmits a short pulse of constant RF frequency that is scattered from the targets and returned to the transducer. The target echoes are mixed with an electronic off-axis reference signal in a manner that is similar to the simultaneous source-receiver system. The mixer output modulates the intensity of a scope, and a transparency is recorded from the scope's face. A one-dimensional hologram is produced by all the point targets that vary in depth into the object, and Figure 11b shows the expected holograms from two such targets. The reconstruction of this collection of holograms is performed by the optical processor (Figure 12) which records on film the reconstructed image. Figure 11c illustrates the processed images of the two point targets described earlier. The complicated optical system (Figure 12), consisting of a conical, cylindrical, and a spherical lens, is required to bring all the holograms into focus at the plane of the photographic film which intercepts the first-order diffraction pattern from these holograms.

The lateral resolution of this system is approximately ten times better than the usual unfocused B-scan system, but the improved resolution is achieved at the expense of complicated electronic and optical processing. This is the price that is usually paid when synthetic aperture techniques are employed. In addition, because these systems are basically holographic, speckle and fringing effects are evident to some extent.

BRAGG IMAGING

Bragg imaging is a technique for the direct visualization of a sound field by diffraction of light. This technique, first developed by Korpel,[24] differs from Schlieren photography in that images are produced by Bragg diffraction rather than from Debye-Sears diffraction. A simplified diagram (Figure 13a) illustrates that a virtual image of an object is formed when light is scattered at the Bragg angle from spherical acoustic wavefronts. The virtual image of an external source is demagnified by the factor Λ/λ, where Λ is the sound wavelengths and λ is the light wavelength. The Bragg angle is very small for most applications and is given by the expression:

$$\sin\theta_b = \lambda/2\Lambda.$$

24. KORPEL, A. *Visualization of the Cross Section of a Sound Beam by Bragg Diffraction of Light.* Appl. Phys. Letters, December 1966, p. 425.

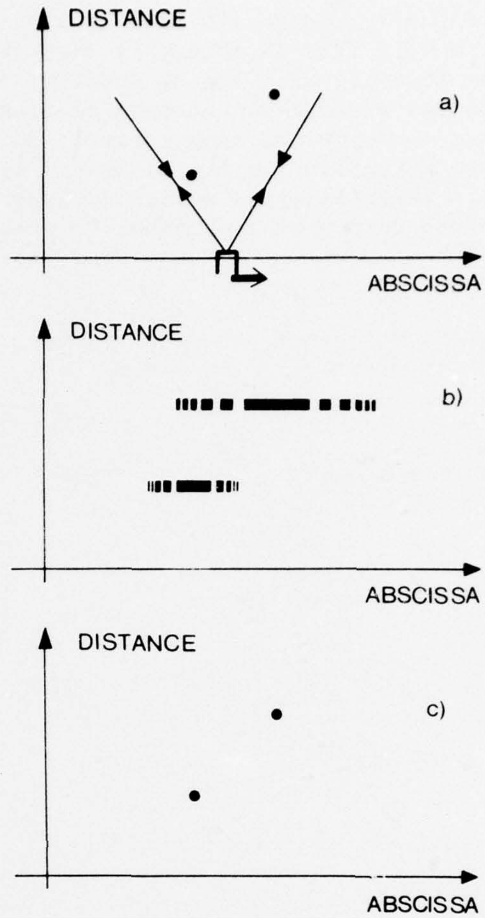


Figure 11. Synthetic aperture sonar: (a) recording arrangement and objects, (b) record, and (c) processed image.

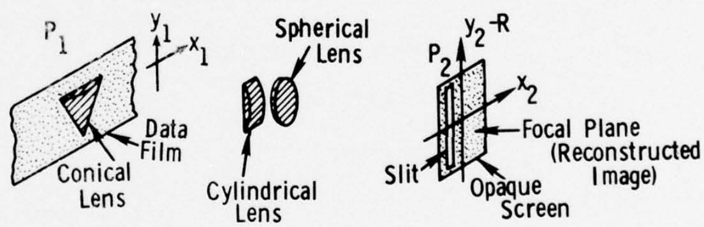


Figure 12. Optical processor. The photographic film is positioned behind the slit at the focal point of the reconstruction lenses, and is exposed to the light passing through the slit.

For example, at 5 MHz the ratio Λ/λ is approximately 500, resulting in a Bragg angle of less than 0.06°. We see that the light illumination must be nearly perpendicular to the direction of the incident acoustic wave. In general, Bragg imaging becomes very difficult at low frequencies because of this small angle, and the best results are obtained with an acoustic frequency higher than 15 MHz.

Recently, Bragg imaging has been applied to NDT applications, at both TRW[25] and the University of California[26] at Santa Barbara. A schematic of the TRW system

25. SMITH, R. A., DOSHI, N. H., JOHNSON, R. L., and BHUTA, P. G. *TRW Acousto-Optical Nonimmersion Flaw-Imaging System.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1973, p. 1.
26. LANDRY, J., KEYANI, H., and WADE, G. *Bragg-Diffraction Imaging: A Potential Technique for Medical Diagnosis and Material Inspection.* Acoustical Holography, G. Wade, ed., Plenum Press, New York, v. 4, 1972, p. 127.

is shown in Figure 13b, and this system is somewhat unusual in that reflected acoustic energy from an object is used for imaging. Normally, a through-transmission mode is employed. Two cylindrical lenses are shown in the figure. The first lens produces a wedge-shaped beam of light that interacts with the scattered acoustic field. The second lens corrects for the demagnification of the image in the transverse direction (in the plane of Figure 13b) and focuses this image onto a TV camera. Essentially, a series of images of the type described in Figure 13a are stacked on top of each other to form a two-dimensional image. This is the reason

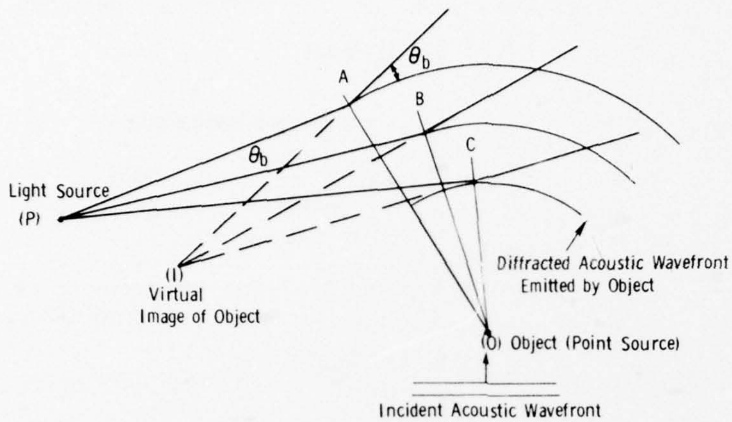


Figure 13a. Bragg imaging.

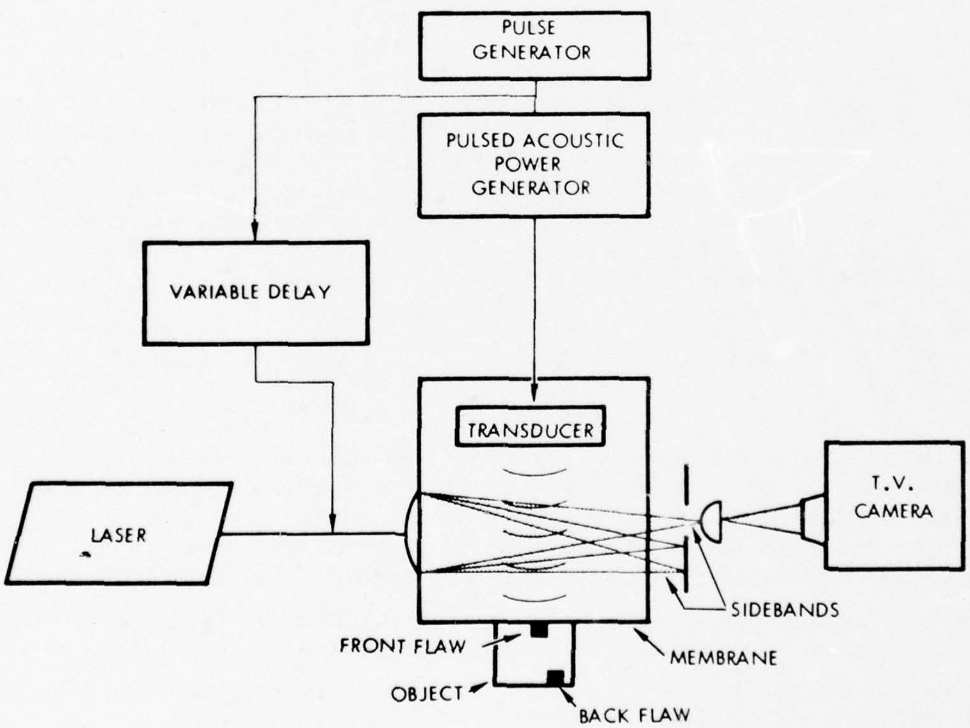
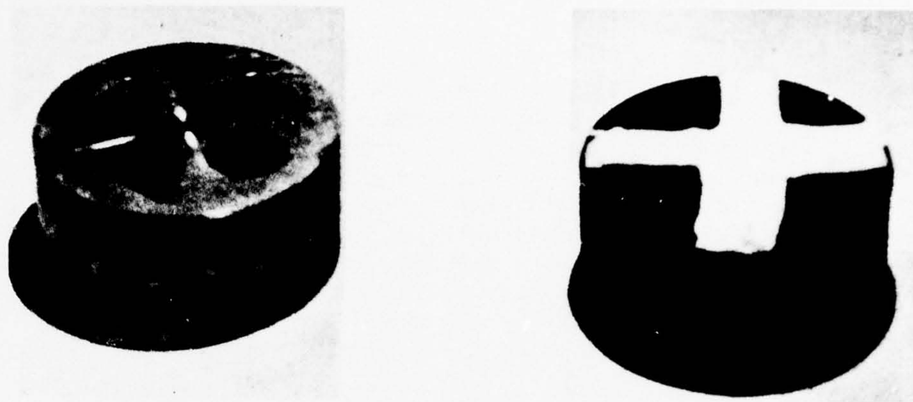


Figure 13b. Block diagram of the TRW Bragg imaging system.

no demagnification of the image occurs in the vertical direction, which is the direction normal to the plane of Figure 13b. Typical images generated by the TRW system are shown in Figure 14.



Test Objects

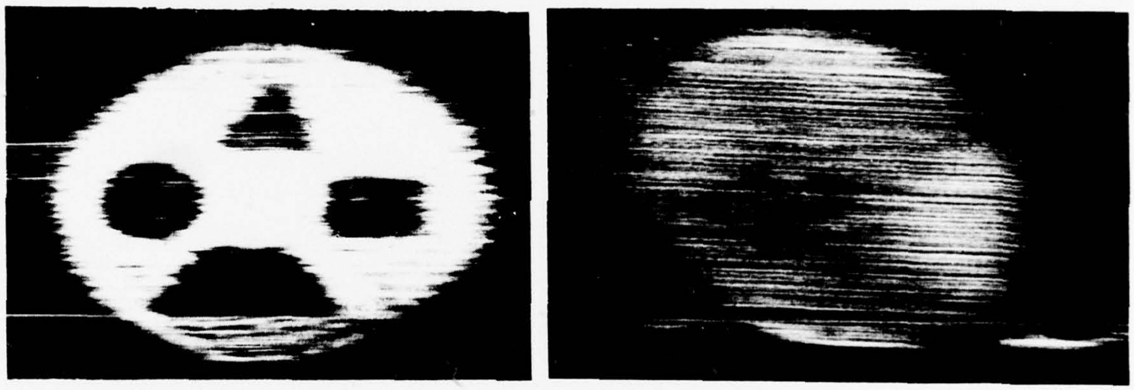


Figure 14. Images from sound reflected within test objects at 18.7 MHz.

The resolution of Bragg imaging systems is much better in the transverse direction than in the vertical direction. Although the transverse resolution may be an acoustic wavelength or so, the vertical resolution is normally ten times this. The sensitivity, at least theoretically, should be about the same as liquid surface holography. However, in practice, it is typically three to five times worse than this.

Bragg imaging is very similar to holography in that phase and amplitude information is retained in the imaging process, although, of course, no reference beam is needed. The same problems associated with speckle and fringing effects are present in the Bragg images because of the highly coherent nature of the imaging process. The strong point with Bragg imaging is that real-time images are formed with a relatively simple and low-cost system. Although the images to date are somewhat poor compared to other techniques, there may be applications where the speed and simplicity of Bragg imaging may be attractive.

RAY TRACING TECHNIQUES

Two-dimensional shadow projections of the absorption of energy through an object can be obtained with light, X rays, acoustic waves, or electrons. Computerized Axial Tomography (CAT) employing X rays has recently made great strides in the field of medical imaging, and several commercial units are now available. The use of this technique for acoustic imaging is more difficult because of reflection and refraction from interfaces having different acoustic impedances. Recently, Greenleaf[27] has done some interesting ultrasonic imaging of biological specimens using this technique, and his work will be briefly discussed here.

Figure 15 illustrates the basic idea behind ray tracing techniques. The image plane is divided into the rectangular grid, as shown, and the acoustic absorption within each square is assumed to be constant. The transmitter and receiver are scanned along a line, and the intensity of the acoustic energy that passes through the object for each particular ray is stored and later analyzed. Each ray generates one equation that contains information about some of the unknown absorption coefficients P_i. The complete set of P_i is needed for an image. The values of the ℓ_{ij} (shown in Figure 15) are known because they can be easily calculated from geometric considerations. Therefore, in order to solve for the n^2 unknown number of P_i, n^2 equations must be generated. This is accomplished by the repeated linear scanning

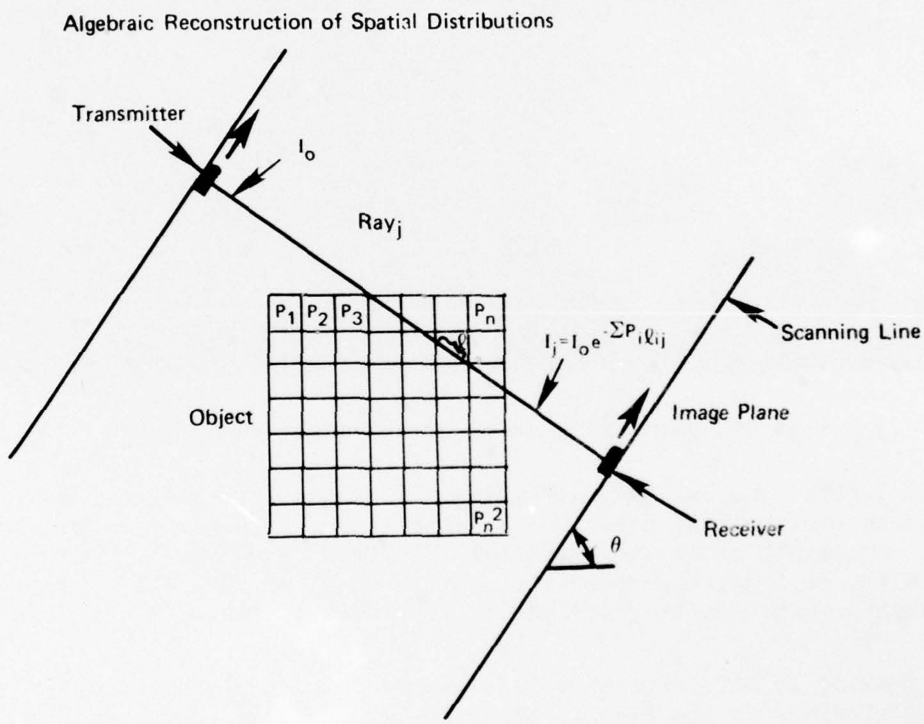


Figure 15. Ray tracing technique.

27. GREENLEAF, J. F., JOHNSON, S. A., LEE, S. L., HERMAN, G. T., and WOOD, E. H. *Algebraic Reconstruction of Spatial Distributions of Acoustic Absorption within Tissue from Their Two-Dimensional Acoustic Projections.* Acoustical Holography, P. Green, ed., Plenum Press, New York, v. 5, 1974, p. 591.

of an object after it has been rotated through a small angle, with the process continuing until a full 360° rotation of the object has been achieved. The computer is programmed to use an algebraic reconstruction technique for solving the large number of simultaneous linear equations.

The images obtained so far with biological specimens have been somewhat crude, primarily because of artifacts introduced by reflection and refraction. As an extreme example, consider the case where the variation in acoustic impedance within the object is so large that the acoustic beam is deflected from a straight line and misses the receiver completely. The received intensity for such a ray would be zero, and the computer would falsely think that strong absorption had occurred. In general then, this type of imaging system will give good results only if the problems of reflection and refraction are not severe in the particular object of interest. This would appear to rule out many of the most common NDT applications.

CONCLUSION

Acoustic imaging has already established itself as a useful tool for medical and NDT applications. The last few years have been a period of very active research; and, as new techniques emerge from the laboratory into practical imaging systems, the impact of acoustic imaging on these two fields will be substantial. The basic principles of acoustic imaging have been known for many years, but it has been the recent revolution in integrated electronics that has led to more sophisticated and less costly systems.

The development of fast scanning linear array systems are particularly important in that they have the capability of greatly reducing the inspection time required for many NDT applications, and they should interface nicely into fully automatic inspection schemes. Real-time medical imaging systems can now be built that monitor fetal activity, determine the plaque content of major arteries, or study heart valve motion. These tools are expected to have a major impact on clinical medicine.

There are still many problems to be solved. Better piezoelectric transducer arrays need to be built. Improvement in sensitivity and resolution is desired, as well as the elimination of speckle and fringing effects. Much work remains to be done and several laboratories are currently addressing themselves to these problems.

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APPENDIX A. REVIEW OF EARLY IMAGING TECHNIQUES

This section will be a brief review of some early methods that were used for converting ultrasonic energy into a useful image. Most of the material discussed here is taken from Reference 1 and this article should be consulted if more information is desired. It is convenient to group the ultrasonic imaging techniques into the following four categories: (a) photographic and chemical, (b) thermal, (c) optical and mechanical, (d) electronic. Table A-1 lists a number of the techniques that fall within these categories, as well as the associated threshold sensitivities.

Table A-1. ULTRASONIC IMAGE DETECTION METHODS

	Approximate threshold sensitivity (W/cm^2)
Photographic and chemical methods	
Direct action on film	1–5
Photographic paper in developer	1.0
	0.1
Starch plate in iodine solution	1
Film in iodine solution	1
Color change effects	0.5–1
Thermal techniques	
Phosphor persistence changes	0.05–0.1
Extinction of luminescence	1
Stimulation of luminescence	—
Thermosensitive color changes	1
Change in photoemission	0.1
Change in electrical conductivity	0.1
Thermocouple and thermistor detectors	0.1
Optical and mechanical methods	
Optical detection of density variations	10^{-3}
	3×10^{-4}
Liquid surface deformation	10^{-3}
	10^{-5}
Solid surface deformation	10^{-6}
Mechanical alignment of flakes in liquid	2.8×10^{-7}
Acoustic birefringence	10^{-1}
Electronic methods	
Piezoelectric detector—mechanical movement of transducer or object, or use of an array of transducers to form an image	10^{-11}
	5×10^{-12}
Probe detection of potential on back of piezoelectric receiver	0.05
Electron scan of piezoelectric receiver	2×10^{-11}
Electron scan of piezoresistive receiver	10^{-7}
Piezoelectric—electroluminescent phosphor detector	10^{-6}

Photographic and Chemical Methods

That ultrasonic energy can affect photographic emulsion was perhaps first reported in 1933 by Marinesco and Truillet. The exact mechanism involved, whether mechanical or thermal, is still unclear; however, the softness of the emulsion does appear to be a critical factor. Useful images usually require a long exposure (several minutes) to relatively high acoustic intensities (1 W/cm^2). Note in Table A-1 that photographic and chemical methods are the least sensitive to acoustic energy in comparison to other techniques.

Thermal Techniques

A variety of thermal techniques have been explored that make use of the heating effects produced by an ultrasonic field. A very simple imaging system might consist of a thermocouple that is scanned across the acoustic field. Compounds such as liquid crystals are also useful for thermal imaging because these materials change colors when exposed to small temperature variations. In addition, a number of temperature-dependent luminescent materials have been used as thermal ultrasonic detectors. However, for most medical and NDT applications, the sensitivity of the thermal techniques is not adequate.

Optical and Mechanical Methods

An example of a successful imaging technique that falls into this category would be a Schlieren optical system. Parallel light (Figure A-1) passing through the water tank is bent by Debye-Sears diffraction as a result of periodic changes in the index of refraction brought about by the acoustic field. Only the diffracted light is imaged at the plane of film because an opaque stop is located at A_2 that intercepts the central, undiffracted light beam. Schlieren photography has been very useful for displaying the acoustic beam profile of piezoelectric transducers (Figure A-2).

Some transparent materials become temporarily birefringent when highly stressed, and this effect can be used for acoustic imaging. Polarized light, when passing through a region of high acoustic intensity, will become partially depolarized. Acoustic images have been made in quartz using sound field intensities of approximately 0.1 W/cm.

An interesting technique, described by Pohlman in 1939, consists of reflecting light from small aluminum flakes that are suspended in a liquid. When no acoustic field is present, the flakes assume a random orientation and very little light is reflected. However, when the acoustic field is applied, the flakes tend to align themselves with the field and strong reflection occurs. This technique is fairly sensitive ($\approx 10^{-7}$ W/cm^2), but the reaction time for the flakes to align themselves can be on the order of a few minutes.

The liquid surface holographic technique, described earlier in this paper, is another example of optical and mechanical methods. This technique was first proposed by Sokolov back in the 1930's, although he had no way of understanding holography in the same sense as we do today. This technique has been included in the section dealing with advanced imaging techniques because modifications made in the last few years have substantially improved the quality of the acoustic images.

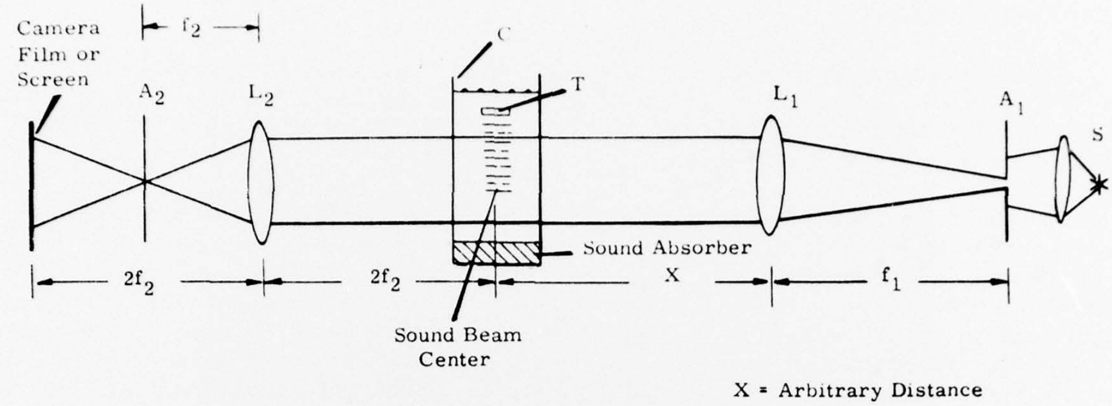


Figure A-1. Diagram of Schlieren apparatus.

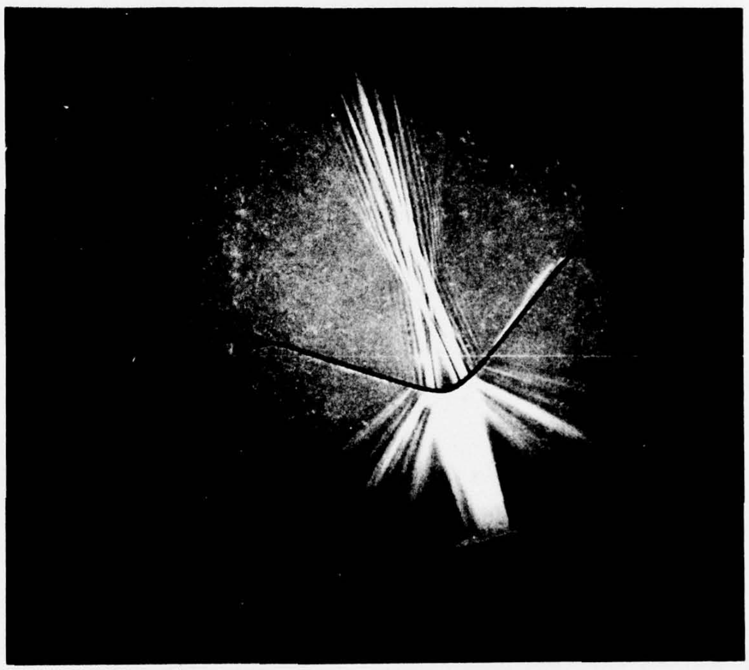


Figure A-2. A Schlieren photograph of a continuous-wave ultrasound beam from a 5-MHz, 1.87-cm diameter, zirconate titanate transducer (beam enters from the right). The RF power was 20 W. The ultrasound beam is shown striking a thin sheet of aluminum. Reflected and transmitted ultrasound beams can be readily observed. (Courtesy of G. J. Posakony)

Electronic Methods

Electronic methods are by far the most sensitive techniques for detecting acoustic energy. Piezoelectric transducers have been used for many years, and nearly all the advanced techniques that are being developed today rely on these transducers as acoustic detectors.

Several advanced techniques based on electronic methods have been summarized, but there is an interesting imaging tube, again developed by Sokolov, that should be mentioned. The Sokolov tube is a sort of ultrasound television camera tube that typically consists of a thin quartz faceplate. When the front of the quartz is exposed to an acoustic field, a high velocity electron beam is made to scan the back of this faceplate. Secondary electrons, modulated by the piezoelectric voltages that are generated by the sound field, are collected, amplified, and converted into a TV image. Because the thin piezoelectric target serves as the tube faceplate, a compromise must be made between resonant frequency and the physical size of the faceplate. In general, only low-frequency imaging (<5 MHz) is possible.

APPENDIX B. ULTRASONIC TRANSDUCERS

This will be an overview of some of the more important properties of ultrasonic crystal transducers. Piezoelectric crystal such as quartz, rochelle salt, and lithium sulfate generate acoustic waves because they are mechanically strained when an electric field is applied in the direction of a polar axis. On the other hand, ceramic materials such as lead zirconate titanate and lead meta-niobate are not truly piezoelectric, but are rather ferroelectrics. These crystals are effective for ultrasonic transduction if the electric dipoles are first polarized in the crystal. When polarized, ceramic transducers display the same "piezoelectric effect" as true piezoelectrics provided the exciting electric field is small compared to the initial polarizing field. In fact, ceramic materials have some advantages over true piezoelectrics in that they are polycrystalline and do not have to be cut along specified crystal axes. They can be machined to a concave surface to easily form a focused transducer.

Electrical Circuit Considerations

A simple ultrasonic transducer is essentially a capacitor with backing material on one of its faces (see Figure B-1). Important parameters, such as frequency response and insertion loss, can be determined from a knowledge of the equivalent circuit of a transducer. A simplified equivalent circuit of a transmitting crystal near resonance is shown in Figure B-2. This circuit gives us information about the performance characteristics of a transducer and how to more efficiently couple energy into the transducer using electrical matching schemes.

The right of the circuit in Figure B-2, consisting of the series circuit of L_1, C_1, and R_1, represents the mechanical properties of the crystal. These circuit elements are functions of the piezoelectric parameters of the crystal, the acoustic impedance of the loading and backing material, and the size and orientation

Z(acoustic impedance) = ρ(density)v(velocity)

t(thickness) of crystal = $\lambda/2$ (at resonance) = $v/2f_0$

C_0(static capacitance) = ϵ A(area)/t

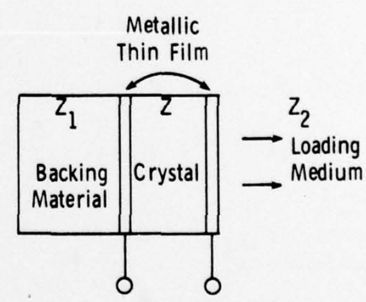


Figure B-1. Geometry of disk transducer.

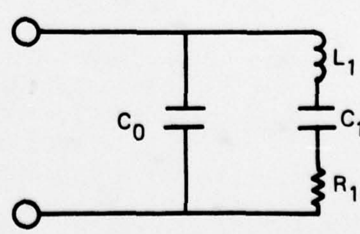


Figure B-2. Equivalent circuit of transmitting transducer near resonance.

of the crystal. References[28,29] should be consulted for a more complete description of these circuit elements. The capacitor C_0 is the static capacitance of the crystal.

As an example of the use of this equivalent circuit, consider the problem of determining the frequency response of the transducer. The best approach would be to solve the complete circuit for the frequency-dependent voltage drop across the load resistor R_1. Maximum acoustic generation occurs when the electrical power into the load resistor R_1 is maximum because this term represents the conversion of electrical energy into acoustic energy. An approximate procedure for determining the frequency response is to divide the Q of the circuit into two parts, Q_{mech} and Q_{elect}. The Q factor of the circuit determines the frequency response for that circuit and efficient operation is possible when Q_{mech} and Q_{elect} are approximately equal.

Q_{mech} represents the piezoelectric and physical parameters of the circuit and this is the Q of the series circuit of L_1, C_1, R_1.

$$Q_{mech} \cong f_0/f_1 - f_2 = (2\pi f_0 L_1)/R_1$$

where $f_0 = 1/2\pi\sqrt{L_1 C_1}$ and f_1 and f_2 are 3 db points.

Q_{elect} represents the effect of adding matching elements to the circuit. For example, if a tuning inductor (L) was connected across the input in order to cancel the reactance of C_0 at resonance, Q_{elect} would be:

$$Q_{elect} = R_1/2\pi f_0 L.$$

Generally, for wide-band operation, it is desirable to design and match a transducer such that both Q_{mech} and Q_{elect} are approximately equal to unity. This often requires that a damping resistor be inserted in the circuit in parallel with the tuning inductor to reduce Q_{elect}. The damping control on most pulser-receivers is simply a variable resistor that connects to the input of the transducer so that Q_{elect} can be varied. When both Q's are matched, the frequency response of the transducer is given approximately by:

$$\Delta f/f \simeq 1/Q_{mech} = 1/Q_{elect}.$$

Another use of the equivalent circuit is to determine the acoustic power density generated by a transducer. For an air-backed $\lambda/2$ thick transducer at resonance:

$$P \text{ (Acoustic Power Density)} \simeq \frac{16\ d^2 z^2 V^2 f_0{}^2}{Z_2}$$

where d = strain developed/applied field
V = voltage applied (rms)
f_0 = resonance frequency
Z = acoustic impedance of crystal
Z_2 = acoustic impedance of loading medium.

28. GOOBERMAN, G. L. *Ultrasonics, Theory and Application.* Hart Publishing Company, New York, 1968.
29. BLITZ, J. *Fundamentals of Ultrasonics.* 2nd ed., Plenum Press, New York, 1967.

A 1-MHz quartz transducer, loaded by water and driven by a 100-volt oscillator, would have an output of 0.01 W/cm^2.

Transducer Beam Profile

1. Unfocused Transducers

The beam profile of an ultrasonic transducer is determined by diffraction effects that are governed primarily by the ratio between transducer crystal size and ultrasonic wavelength. It is convenient to divide the profile pattern into two zones: (1) the Fraunhofer zone, or far field, and (2) the Fresnel zone, or near field. Figure B-3 shows the relative acoustic intensity of a circular transducer along an axial line as a function of distance away from the crystal. The intensity in the near field is strongly distance dependent, and this should be kept in mind when contact ultrasonic testing is used. For most immersion testing, the object is exposed to a far field pattern, where the beam is well defined and not affected by the kind of undulations found in the near field. The dividing line between the two zones is referred to as the Y_0^+ point:

$$Y_0^+ = a^2/\lambda$$

where a = the radius of the crystal
λ = the wavelength of sound in the medium.

Perhaps Figure B-4 gives a better indication of the way the acoustic profile varies with distance away from the transducer. Although beam spread is not indicated in this figure, it should be pointed out that the beam begins to noticeably spread in diameter in the far zone. The angular width of the beam in the far zone is given by the expression:

$$\theta = \sin^{-1}(1.2(\frac{\lambda}{2a})).$$

The results so far pertain to a circular transducer that is driven at a single frequency. What happens if a broad band of energy is excited, as is often the case for pulse-echo testing? The problem is a difficult one, but in general the rapid spatial variations in the intensity that usually occur in the near zone are washed out somewhat. It is still a good idea to avoid testing in the near zone, but at least the acoustic beam profile is better behaved when broad-band operation is used.

2. Focused Transducers

The effect of a lens, placed in front of a planar transducer, is to move the characteristic near-zone patterns (Y_0^+, Y_1^-, Y_2^-) closer to the transducer. The shape of these patterns is essentially unchanged, and they are simply moved in toward the crystal. A properly designed acoustic lens will produce a highly concentrated beam at the focal point. This point is actually the Y_0^+ point, the boundary line between the near and far fields for the focused transducer. Figure B-5 shows a transducer with a simple spherical lens. If we ignore aberrations and consider the small aperture case, then the focal distance is given approximately by the equation:

$$f \text{ (focal distance)} = R(n/n-1)$$

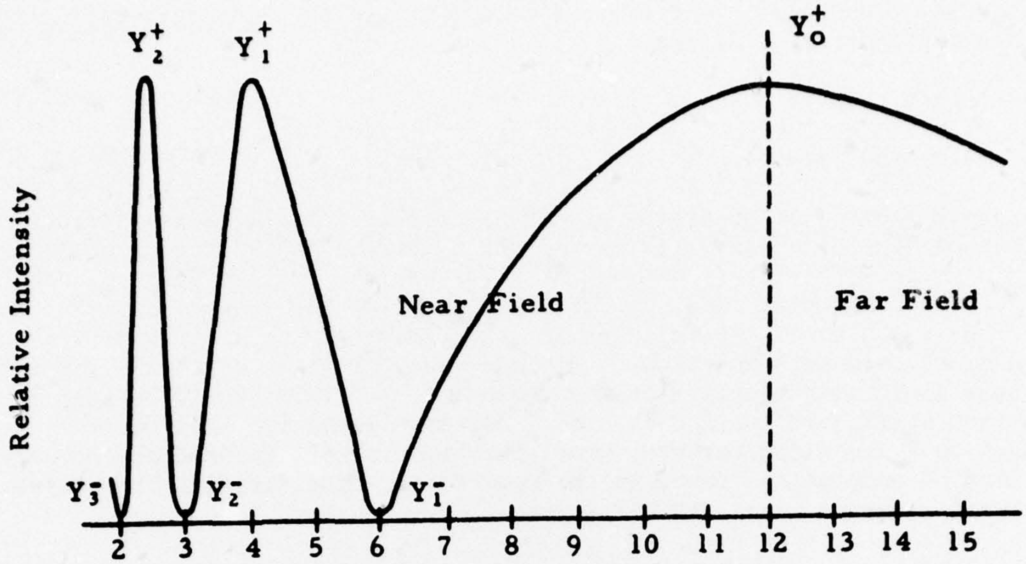


Figure B-3. Axial pressure distribution. (Courtesy of J. T. McElroy)

Figure B-4. Composite view of axial and profile beam intensity. (Courtesy of J. T. McElroy)

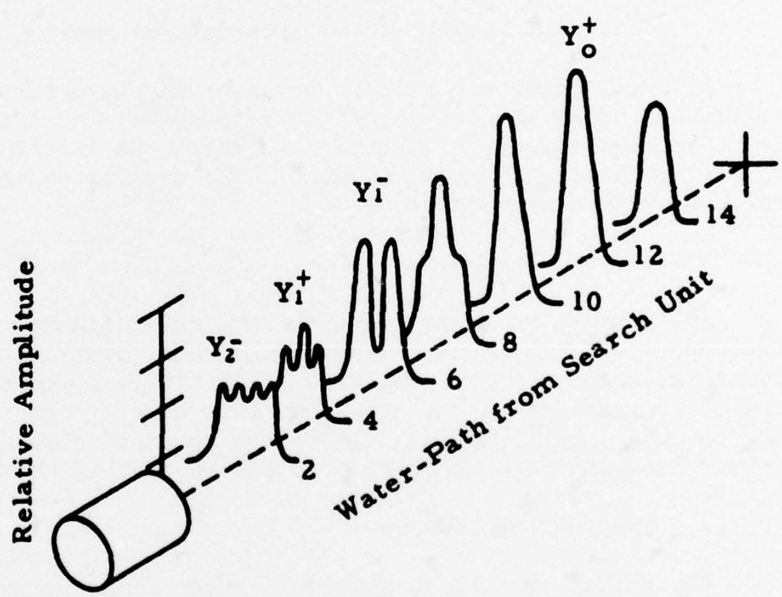
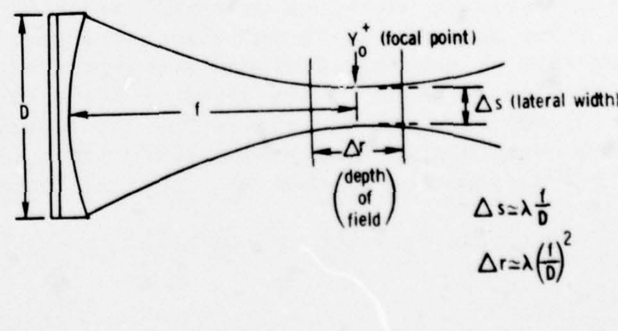


Figure B-5. Focused transducer.



where R = lens radius of curvature
n = index of refraction of lens (relative to water).

The lateral width Δs and the depth of field Δr are also given in the figure. The lateral width of the beam is particularly important for this will primarily determine how "sharp" an acoustic image can be in the direction perpendicular to the axis of the transducer. We have considered here only the case where the beam width is determined by diffraction effects. Aberrations will tend to produce a larger beam width.

Transducer Arrays

Since most of the fast-scanning acoustic imaging systems require arrays of piezoelectric elements, it is important to consider some of the design considerations that must be kept in mind when these arrays are used. Typically, a linear array will consist of a large number of piezoelectric elements which are coated with a conducting film and bonded to a lossy, impedance-matched backing material. This backing absorbs the acoustic energy that would propagate in back of the elements and favorably affects the input electrical impedance of the elements so that broad-band operation is more efficient. In addition, this backing should be conducting in order to provide a good ground plane.

Because it is difficult to fabricate an array consisting of a large number of individual elements, a popular approach is to construct a slotted array. Here, a slab of piezoelectric material, bonded to the appropriate backing, is cut into slots to produce the necessary elements. The saw cuts do not go completely through the piezoelectric slab, but they are deep enough so that cross-coupling between the elements is minimized.

The linear arrays must meet tough design characteristics. The elements must have a high electromechanical conversion efficiency and should be capable of broad bandwidth operation with a relative bandwidth of at least 50%. In addition, the cross-coupling between elements must be small and the elements precisely machined to close dimensional tolerances. Of course, the ease and cost of fabrication is also an important consideration, especially if mass production is anticipated.

It is interesting to consider the broadside far-field pattern from a linear array of equally spaced elements. A knowledge of this pattern is very important, particularly for phased array systems. As a simple example, let us assume that each transducer element radiates uniformly in all directions. We know that real transducers do not radiate this way, but this assumption will not change the important conclusions that we hope to illustrate.

The radiation amplitude at point P for isotropic radiations is given by the approximate expression (see Figure B-6):

$$A(\xi,R) = K(R)\sum_{n=0}^{N-1} \exp(i(2\pi n/\lambda)\sin\xi)$$

where ξ is the angle defined in the figure and R is the distance from point P to the origin.

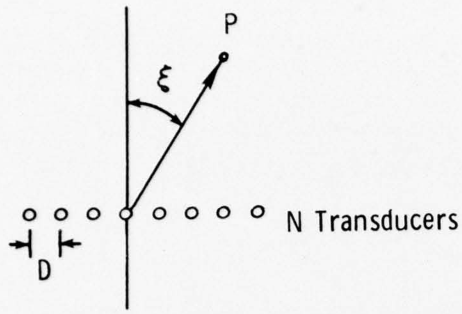


Figure B-6. Linear array geometry.

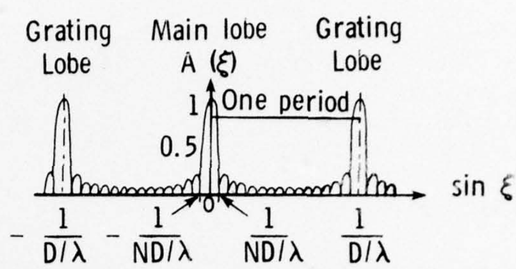


Figure B-7. Radiation amplitude as a function of angle.

A plot of this amplitude as a function of angle is given in Figure B-7. We note two important things. First, the angular beamwidth of the main lobe is approximately equal to $\sin^{-1}\lambda/ND$. Since ND is equal to the length of the array, the beamwidth or lateral resolution of the array is determined by the ratio of the acoustic wavelength to the length of the array. The longer the array, the better the lateral resolution. Second, we note the presence of grating lobes. These lobes can lead to ambiguous information because the array is not only sensitive to acoustic energy normal to the array ($\xi = 0$), but also to energy at the angles $\pm\sin^{-1}\lambda/D$. However, if the spacing D is chosen to equal $\lambda/2$, the grating lobes will appear at $\pm 90°$ and will no longer be a problem. The requirement that the spacing between the elements be less than half the acoustic wavelength follows also from the famous "Sampling Theorem". It is important, therefore, that the spacing between the elements be properly chosen in order to avoid these grating lobes.

What about real transducer elements that do not radiate energy uniformly in all directions?

The angular dependence of the transducer reduces the grating lobe level. But, in general, grating lobes are still a serious problem, and the element spacing considerations still apply.

APPENDIX C. RANGE (AXIAL) RESOLUTION AND SENSITIVITY CONSIDERATIONS IN PULSE-ECHO SYSTEMS

The term "range resolution" implies the ability to determine accurately how far a target is located from the transmitting transducer. A pulse-echo approach is potentially superior to through-transmission with regard to range resolution, because the time duration of the transmitted acoustic pulse can be made short enough so that the range resolution is determined by the pulse length and not by the focusing properties of the transducer. In an ordinary pulse-echo system, in which the transducer is excited by a voltage spike that lasts for a very short time interval, the shorter the pulse, the closer two small targets may be together before their echoes begin to overlap. If the system was completely free from background noise, and the shape of the transmitted pulses were precisely known, then the echoes from two point targets could be distinguished no matter how close together they were. However, in practice it is not feasible to separate echoes when they begin to significantly overlap each other, and the range resolution then depends on the time duration (Δt) of the transmitted pulse.

Now the distance R to a target is determined by the equation:

R (Range Distance) = $tc/2$

where t is the round trip time for the acoustic pulse and c is the acoustic velocity. Therefore, it follows that:

ΔR (Range Resolution) = $\Delta tc/2$.

The range resolution equation can be related to the bandwidth B of the transmitted pulse by taking the Fourier transform of the transmitted pulse. It follows that:

$$\Delta t \simeq 1/B.$$

Therefore, $\Delta R \simeq c/2B$.

This equation is very important because it tells us that broad-band pulses are needed to achieve good range resolution. This is the reason broad-band, highly damped transducers are used when good range data is required.

The term "sensitivity" implies the ability to detect small targets that may be buried in background noise. The ability to detect a target depends on several factors such as the size, geometry, and acoustic impedance of the target, the distance between the target and the transducer, the transmitted acoustic energy, and the level and type of noise in the system.

As an example of how the sentivity of a system might be evaluated, consider the problem of detecting flaws on the inner surface of a nuclear pressure vessel using an immersion pulse-echo technique from the outside. The steel is 28 cm thick and contains cladding on the inner surface. First, the thermal noise from the 1"-diameter unfocused transducer and receiver is evaluated.

Thermal Noise Considerations

Thermal noise power (for 1 MHz bandwidth) = KTB = 4×10^{-15} W

Thermal noise power density	$= 8.1\times10^{-16}$ W/cm^2 → -151 db (ref. to 1 W/cm^2)	
Receiver noise figure	= +3 db	
Total noise (ref. to 1 W/cm^2)	= -148 db	

Next, the losses of the system are computed. Assume the transmitter produces an output level of 1 W/cm^2.

Losses of System	db
Receiver transducer efficiency (for Q ≃ 1)	-7
Transmission twice through water/steel interface	-20
Grain scatter is cladding, round trip (1.6 db/mm at 2 MHz)	-39
Grain scatter in steel wall, round trip (0.03 db/mm at 2 MHz)	-20
Reflectivity of critical defect (energy reflected to receiving transducer)	-30
Total Losses	-116 (ref. to 1 W/cm^2)

The calculation for the reflectivity of the critical defect is extremely difficult. For simple flaw geometries, the scattering cross section can sometimes be evaluated as well as the total energy scattered back to the receiving transducer. In this example, the value of -30 db was arbitrarily chosen. A discussion of how one arrives at the size, shape, and acoustic properties of a critical defect for a particular material is well beyond the scope of this report.

The signal-to-noise ratio for this example would be:

$$\frac{S}{N} = \frac{\text{defect echo}}{\text{thermal noise level}} = -116 \text{ db} - (-148 \text{ db}) = 32 \text{ db}.$$

From this calculation, the system would appear to provide more than enough sensitivity to detect a critical flaw. However, we have assumed that the principal source of noise was thermal fluctuations in the transducer and receiver. In many NDT applications, an A-scan display will show a grass-like background noise level that may be caused, for example, by reflections from small grain boundaries in the material. If the grass level exceeds the thermal noise level, then the signal-to-noise ratio is limited by this source of "noise".

Coherent "grass" noise presents a formidable problem because the improvement of the signal-to-noise cannot be accomplished by standard signal processing techniques. For those applications where thermal noise does indeed dominate, matched filtering and correlation schemes have proven to significantly increase the sensitivity without sacrificing resolution. These techniques were first developed for radar applications and have found only very limited use in ultrasonics. A discussion of these signal processing schemes is beyond the scope of this report, but references[30,31] are cited that may be useful for those interested in pursuing this subject further.

30. BERKOWITZ, R. S. *Modern Radar, Analysis, Evaluation, and System Design.* John Wiley and Sons, New York, 1965.
31. SKOLNIK, M. I. *Introduction to Radar Systems.* McGraw-Hill Book Company, 1962.

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