

ULTRASOUND TRANSMISSION IN TISSUE VISUALIZATION

William J. Fry, Francis J. Fry, Elizabeth Kelly, Thomas A. Fry, & Gene H. Leichner*

INTERSCIENCE RESEARCH INSTITUTE
CHAMPAIGN, ILLINOIS

INTRODUCTION

The existence of differences in the characteristic acoustic impedance of adjoining tissues of different types is critical for the operation of ultrasonic visualization systems which depend on partial reflection of acoustic energy at tissue interfaces for detection of such structural features. At normal incidence, reflection of acoustic energy will take place only if a difference in impedance exists at an interface. Table 1 shows values of the speed of sound and the characteristic acoustic impedance (product of speed of sound and density) for various tissues and physiological saline. The difference between the values of the acoustic impedance for most soft tissues is a relatively small percentage of the tissue impedance so that the fraction of incident energy reflected at a tissue interface is consequently small. Nonetheless, sophisticated transducer and electronics design has enabled small amplitude reflections to be detected with the result that present ultrasonic visualization systems are capable of revealing structural features not observable in vivo by any other means. Although investigators may continue to improve systems based on the indicated principle by further advances in instrumentation, the small differences in acoustic impedance between some considerably different but adjacent tissues limits the degree of improvement that can be achieved. Ideally, it would be desirable to increase the amplitude of the fraction of incident energy that is reflected by enhancing the difference in acoustic impedance between adjacent structures. One could approach this problem by considering the direct application of chemical agents to affect tissue density or the application of hydration and dehydration techniques (Baum, 1965), but these approaches, although valuable in certain restricted cases, have obvious limitations. The technique suggested by Fry (1962) has the advantages that the acoustic impedance change can be induced without physical interference with the tissue, the effect is rapidly reversible, it can be restricted to small volumes if desired, and it can be applied to the examination of a wide variety of tissue structures (Fry and Fry, 1963; Fry, 1965).

The Fry method is based on the fact that in general the value of the acoustic absorption coefficient is a more sensitive indicator of changes in the structure of both biological and physical materials than is the acoustic impedance. (See, for example, Mason and McSkimin, 1947; Roderick and Truell, 1952; Carstensen et. al., 1953; Espinola and Waterman, 1958; Auberger and Rinehart, 1961; Fry and

* The present research is supported under grant (CA 07043). Some of the research discussed in the paper was previously supported under a Bureau of Ships Contract Nobs-88637 and an Office of Naval Research Contract Nonr-4800(00)(x).

Dunn, 1962; Dunn, 1965.) That is, relatively large differences in the value of this parameter can exist for tissues in cases where the percentage difference in the values of the acoustic impedance for the same tissues is quite small. Consider, for example, the gray and white matter of the brain. The values of the acoustic impedance are apparently so close that boundaries between these two tissues cannot be located by present ultrasonic visualization techniques, but acoustic absorption coefficient values for dense gray and myelinated tracts (white) amount to a factor of almost two (Barnard et. al., 1955). In the method proposed by Fry, such differences in acoustic absorbing characteristics permit the detection of neighboring or mutually imbedded structures by the following technique: One pulsed beam of ultrasound is used to produce, by conversion of acoustic energy to heat, transient temperature changes within the tissue under examination. The use of a focused beam presents some advantages for this purpose and when the focus includes a portion of the boundary region between tissues characterized by different values of the absorption coefficient, gradients of temperature at the mutual interface would be produced because of the change in the absorption coefficient at the interface. Since the speed of sound in tissue changes with the temperature, such heated boundary regions between structures of various types would be characterized by gradients in the speed of sound, and these gradients can be detected by normal reflection techniques.* Only a small change in the speed of sound need be produced in order to cause a detectable amount of ultrasonic energy to be reflected when the examining ultrasound beam is used to scan the region of the focus of the temperature inducing beam. For tissue interfaces where the change in absorption is large enough no damage to the tissue structure would result since the temperature changes induced could then be restricted to only a few degrees. (Calculations in the subsequent part of this paper serve as an example.) The examining or detecting beam would be synchronized with the heating pulse so that its train of short pulses would be focused into the region of the induced temperature change just succeeding and possibly during the terminating phase of the relatively long duration pulse of the temperature inducing beam.

TABLE I
SPEED OF SOUND AND ACOUSTIC IMPEDANCE

Species	Tissue	Condition	Temp. °C	Frequency Mc	Velocity m/sec	Specific acoustic impedance g/cm ² x sec x 10 ⁵
Man	Limb (mostly muscle)	In vivo	Body	2.5	1540 (mean value)	1.63
Man	Liver	Refrig.	24	1.8	1570	
Man	Fat	Refrig.	24	1.8	1476	
Man	Skull bone	Fresh	Body	0.8	3360	
Dog	Brain	Fresh	24	2.5	1515	1.56
Dog	Muscle	Fresh	26	4.12	1592	
Hog	Brain		24	2.5	1506	1.55
Castor Oil			25		1466.5	1.395
Physiological saline (0.9% NaCl)			25		1515.3	1.520

* Gradients would also be produced in tissue homogeneous with respect to acoustic absorbing properties but the magnitudes of these (important from the viewpoint of producing artifacts detected by examining pulses) compared to those at the tissue interfaces would be controlled by shaping of the focus of the heating beam.

THEORETICAL ANALYSIS

An analytical evaluation of the required values of the acoustic parameters to implement the method outlined above indicates that it is feasible. Parameters of significance are: the duration and intensity of the heating pulse, the amplitude of the temperature rise in the tissue caused by the heating beam, the temperature coefficient of acoustic impedance, and the operating frequencies of the examining and heating beams. It is convenient to calculate first the effect of the duration of the heating beam on limiting the resolution. The structural resolution that can be achieved by the method of interest here is dependent upon the steepness of the temperature gradients, and consequently, gradients in sound speed at the tissue interfaces produced in the region of the focus of the temperature modifying pulse. The steepness of this gradient per unit temperature difference, between two different types of tissue characterized by different values of the absorption coefficient, is dependent on the coefficient of thermal conductivity. The greater the conductivity coefficient the more rapidly thermal equilibrium is approached. In the calculation of heat conductivity relations the basic physical parameters of interest are: density, ρ , heat capacity per unit mass, s , thermal conductivity coefficient, k , and a quantity, the diffusivity, a^2 , which is derived from these parameters--i.e., the ratio of the thermal conductivity coefficient to the heat capacity per unit volume. For the purpose of the present evaluation, water may be used as the material for calculating the values of these parameters. In order to simplify the calculation of the spatial distribution of the temperature in the tissue as a result of applying the heating beam, it is assumed that the temperature can be approximately represented by the distribution which occurs at a time equal to the heating pulse duration after the instantaneous application of a simple temperature step function. The temperature distribution, T_1 , following the placement of a one-dimensional unit temperature step function in the tissue, and centered at the plane $x = 0$, is*:

$$T_1 = 1/2 \left\{ 1 + \phi \left(\frac{x}{2a\sqrt{t}} \right) \right\}$$

where the function ϕ is defined by the expression

$$\phi(u) = \frac{2}{\sqrt{\pi}} \int_0^u e^{-\xi^2} d\xi, \text{ AND } a^2 = k/\rho s, \text{ **}$$

x is the space coordinate, t designates time, and a^2 is the diffusivity. If one takes, for convenience, the value of a^2 as equal to $0.001 \text{ cm}^2/\text{sec}$ (the value of a^2 for water is 0.0014), then the normalized temperature distribution is illustrated in Fig. 1 for various time intervals measured in seconds after the application of the temperature step function. The position $x = 0$ may be considered as the boundary between two tissues characterized by different absorption coefficient values and as the surface at which the focus of the heating beam is centered. Therefore, the values of resolution correspond to increments of x . For the purpose of estimating resolution perpendicular to the interface the portion of the curve of greatest steepness corresponding to one half of the amplitude of the temperature step is considered. The duration of the pulse that can be employed to induce the temperature distribution in the tissue is then chosen on the basis that it should be less than (but of the same order as) the time interval, after initiation of the temperature step, which causes the central half amplitude of the step to lie within the predetermined space resolution interval. For example, for a resolution of 0.2 mm , a pulse duration of the order of 0.1 sec is indicated.

The determination of an appropriate intensity for the heating pulse is based on two considerations, namely, (1) a transient temperature rise in the tissue which is not injurious, and (2) a fractional acoustic impedance change in the transition region of the structures to be visualized which is of sufficient magnitude that a detectable acoustic reflection will be received when the region is scanned by an examining beam. The percentage change in impedance is determined

* See for example Section 7.14 in H. Margenau and G. M. Murphy, *The Mathematics of Physics and Chemistry*, New York, O. Van Nostrand Co., Inc., 1943.

** At $t = 0$, $T_1 = 0$ for $x < 0$ and $T_1 = 1$ for $x > 0$.

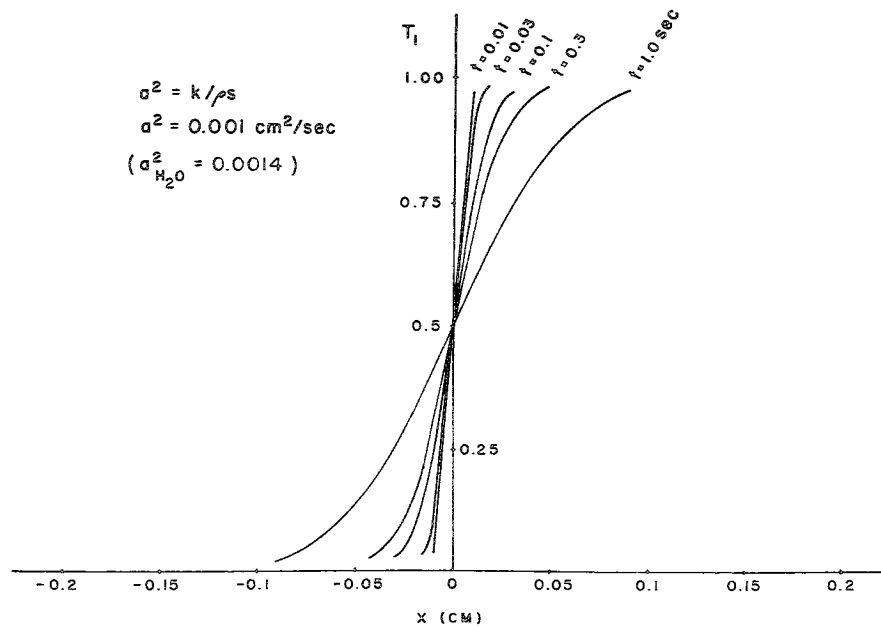


Fig. 1. Temperature distribution as a function of distance from the plane $X=0$ for various time intervals after induction of a unit temperature step function centered at this plane.

primarily by the change in sound speed induced by the temperature increase since the temperature coefficient for density is usually considerably smaller than that for sound speed. In calculating the intensity of the heating beam to be used in the visualization of brain structure, the calculations of the present paper are based on the restriction that there be a safety factor of 10 between the sound energy required to produce a histologically observable lesion in the tissue, and the energy required to produce a temperature gradient appropriate for the required differential impedance change. Previous experimental studies of the present authors have indicated that for a frequency of 1 Mc, an intensity of approximately 1000 w/cm² and a time duration of 1.00 sec, a discernible lesion is produced in dense white matter of brain tissue. Now the maximum temperature rise in the tissue is given approximately by:

$$T = \frac{\mu I t}{\rho C}$$

where the value for the absorption coefficient, μ , is taken as 0.2 cm⁻¹ (average for brain tissue at 1 Mc), and I , t , and ρC designate respectively intensity in w/cm, time in sec, and heat capacity/unit volume in joules/cm³. Therefore, for a frequency of 1 Mc, an intensity of 1000 w/cm², and a time duration of 0.1 sec, the maximum temperature rise is approximately 5°C, resulting in a temperature increment between white and gray matter of approximately 2°C. A temperature rise in brain of the order indicated for such a brief period of time would not appear to warrant concern in view of the "safety factor" already indicated, but the matter will receive further attention before diagnostic application to the human is contemplated. In the preliminary studies a resolution of the order of 0.5 to 1.0 mm, with a consequent reduction in the intensity of the beam and a prolongation of the duration, would be attempted rather than trying to achieve immediately a resolution of 0.2 mm.

A temperature difference of the above magnitude results in approximately 1/4% change in the speed of sound assuming that the temperature coefficient for the speed is equal that of Ringer's solution. The ratio of reflected to incident intensity at the interface is thus $(10)^{-6}$, and if a further reduction of $(10)^2$ occurs due to beam spreading (the use of a collector of large solid angle is assumed) for the returning echo and a reduction of a factor of 5 caused by absorption also occurs, then the intensity at the detector is $(1/5)(10)^{-5}$ w/cm². This corresponds to a pressure amplitude of 1.4×10^3 dynes/cm². Since a probe sensitivity of 10^{-7} volts/dyne/cm² is readily achieved the amplitude of the returning echo would correspond to a voltage level across the probe of approximately 140μ volts.

In deciding on a choice of the frequency for the examining beam, a consideration of primary importance is that the reflection coefficient at the interface should be close to that characteristic of an abrupt change in impedance. To approach this problem, one can compare the reflection coefficient characteristic of a step discontinuity to that characteristic of a continuous transition in order to determine under what conditions the continuous transition case would provide an amplitude of reflection of acoustic energy comparable to the step discontinuity case. To evaluate the effect of a continuous transition on the fraction of incident energy reflected, it is convenient to approximate the velocity function in the transition zone by a simple quadratic form, that is

$$V_2 = V_1 \left[1 + \left(\frac{\Delta V}{V_1} \right)^2 \left(\frac{x}{l} \right)^2 \right],$$

where the symbols are explained by the accompanying figure and where

Medium	Transition Region	Medium
1	2	3
v_1	v_2	v_3

$$\frac{\Delta V}{V_1} \ll 1$$

$\Delta V = V_3 - V_1$. It is assumed that V_1 The magnitude of ΔV is calculated as previously indicated from the temperature difference induced between media (1) and (3) and the temperature coefficient of the speed of sound. The variation of the density in the transition zone is neglected since its variation with temperature is considerably smaller than the variation of the sound speed.

Since the fraction of the incident energy reflected is small, an analysis of the propagation of a plane wave of sound at normal incidence from medium (1) can be carried out by assuming that functions describing the spatial distributions of pressure and particle velocity in the transition zone deviate only slightly from the form characteristic of a medium with a uniform value of the speed of sound. That is, the pressure amplitude in medium (2) is taken as

$$P_2 = P_{2c} \left[\cos \frac{2\pi x}{\lambda_1} + F \left(\frac{2\pi x}{\lambda_3} \right) \right] + P_{2s} \left[\sin \frac{2\pi x}{\lambda_1} + G \left(\frac{2\pi x}{\lambda_3} \right) \right]$$

where the functions F and G represent the deviations from the form characteristic of a medium with uniform velocity. The functions F and G can then be evaluated in series form by utilizing the acoustic wave equation. The fraction of incident acoustic energy reflected can then be expressed approximately as follows:

$$\left(\frac{P_{1-}}{P_{1+}} \right)^2 = \frac{(b-l)^2 + 2b^2 \delta^2 (b-l) \left[G'_0 \cos \frac{2\pi l}{\lambda_1} - F'_0 \sin \frac{2\pi l}{\lambda_1} - G_0 \sin \frac{2\pi l}{\lambda_1} - F_0 \cos \frac{2\pi l}{\lambda_1} \right]}{(b+l)^2 + 2b^2 \delta^2 (b+l) \left[G'_0 \cos \frac{2\pi l}{\lambda_1} - F'_0 \sin \frac{2\pi l}{\lambda_1} + G_0 \sin \frac{2\pi l}{\lambda_1} + F_0 \cos \frac{2\pi l}{\lambda_1} \right]}$$

where

$$b = \lambda_3 / \lambda_1, \quad \delta^2 = \left(1/2\pi\right)^2 \left(\frac{\Delta\lambda}{l}\right)^2, \quad \Delta\lambda = \lambda_3 - \lambda_1,$$

$$F_0 = F/a^2 \delta^2, \quad G_0 = G/a^2 \delta^2$$

and F_0' and G_0' are the derivatives of the functions F_0 and G_0 . Evaluation of this expression shows, for example, that for $b = 1.01$, corresponding to a 1% change in acoustic velocity, a transition zone of $1/4$ wavelength or less reflects practically the same fraction of the acoustic energy incident on it as a discontinuity of the same sound speed difference. Therefore, it can be assumed that the transition zones between the regions of differing absorption coefficients act like a discontinuity in reflecting the acoustic energy, provided the transition zone is no more than $1/4$ wavelength. If, for example, the desired resolution is chosen as 0.2 mm, then $\frac{\lambda}{4} \approx 0.2$ mm represents an appropriate choice, and $\lambda = 0.8$ mm. If the veloc-

ity of sound in tissue is taken as 1.5×10^6 mm/sec, the operating frequency would be approximately 2 Mc.

EXPERIMENTAL ANALYSIS

Experimental results bearing on the method outlined in this paper have been obtained using both biological tissue (brain) and physical material (lucite). The early biological experiments (Fry and Fry, 1963) will not be discussed here except to indicate briefly the general type of experiment and the results obtained. The arrangement which was employed is shown schematically in Fig. 2; it included a cat brain with a portion of the skull bone removed, degassed saline as the transmitting medium, a transducer operating at a frequency of 1.0 Mc/sec and a pulse duration of 1 sec to induce a temperature gradient in a deep brain site, and a standard commercial pulse-reflection unit (unfocused beam, operating frequency 5 Mc/sec). It should be noted that the axes of the inducing and examining beams were oriented approximately mutually perpendicular and individually at about 45° with respect to the horizontal plane. Fig. 3 (A) shows the echo pattern received in the absence of

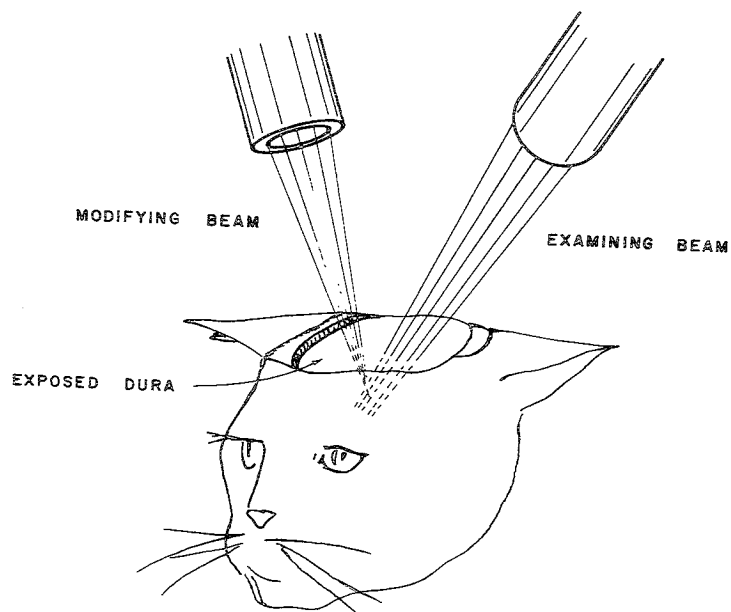


Fig. 2. Configuration of equipment for demonstrating ultrasonic detection of ultrasonically induced acoustic-impedance gradients in brain of experimental animal.

an induced temperature gradient--reflected energy is received from the brain surface at the port of entry (1) of the sound and from the surface diametrically opposite the port of entry (2) where the brain abuts intact cranial bone. A single intervening echo (3) which was just detectable under the operating conditions employed, also appears on the trace. The echo pattern seen approximately two seconds after termination of the pulse that induces the localized temperature gradient is shown in Fig. 3 (B). A large enhancement (factor of 8) of the just-detectable echo of Fig. 3 (A) is apparent (3), and echo (4) from a structural site not exhibited in Fig. 3 (A) is present. Between five and six seconds after generation of the temperature gradient, both the enhanced (3) and the induced (4) echoes are present at reduced amplitude (Fig. 3 (C)), and $1\frac{1}{2}$ min later Fig. 3 (D) demonstrates that the echo pattern had returned to its original form.

The primary objective of the research with the physical specimens was the demonstration that structural inhomogeneities, which are undetectable by the standard acoustic techniques because they exhibit essentially the same values of sound speed and density as the surroundings, may be characterized by considerably different values for the ultrasonic absorption coefficient than those of the imbedding surroundings, and that this difference in the value of the ultrasonic absorption coefficient may be utilized to detect the presence of the inhomogeneity by the method discussed above. It is of interest in regard to the problem of detection of inhomogeneities of structure to note, as previously indicated, that in general the value of the ultrasonic absorption coefficient is more markedly dependent upon the structure of the medium than is the value of the speed of sound. For example, it has been shown that the ultrasonic absorption coefficient changes with the state of strain within a metal (Breazeale and Thompson, 1963). The microstructure of metals is important in determining the magnitude of acoustic attenuation, the grain size being a significant factor. Rayleigh scattering, which is an important ultrasonic attenuation mechanism in solids, is dependent on the presence and orientation of grains and flaws. (For references to pertinent literature, see

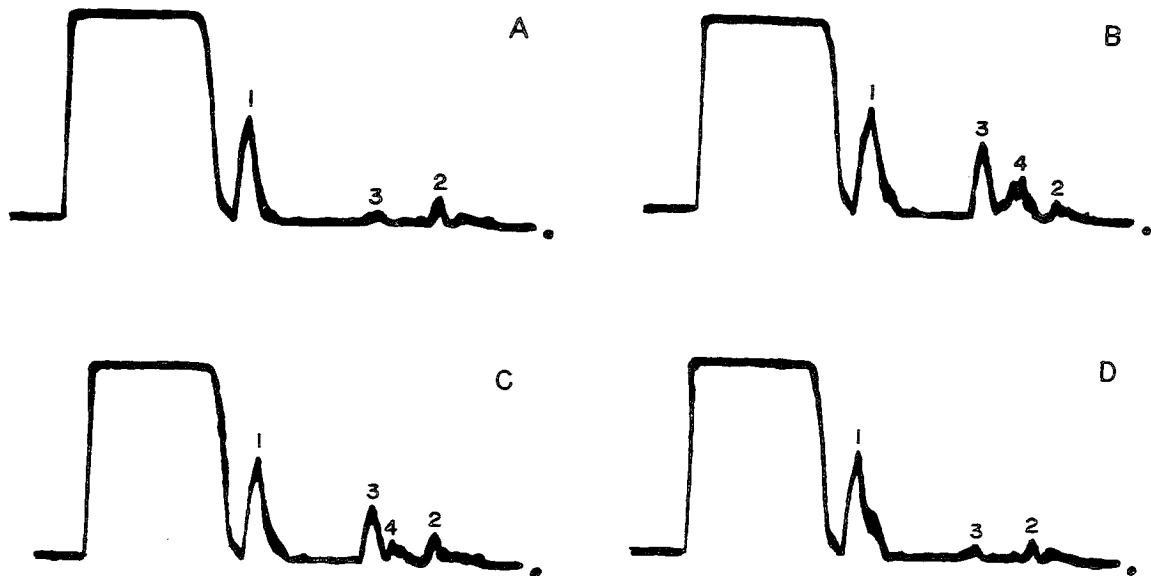


Fig. 3. Echo pattern received by examining transducer, (a) preceding, (b) 2 sec following, (c) 5-6 sec following, and (d) $1\frac{1}{2}$ min. after the induction of a transient-temperature distribution deep in cat brain by a 1.0-sec pulse of focused ultrasound. (Figure is trace of photograph.)

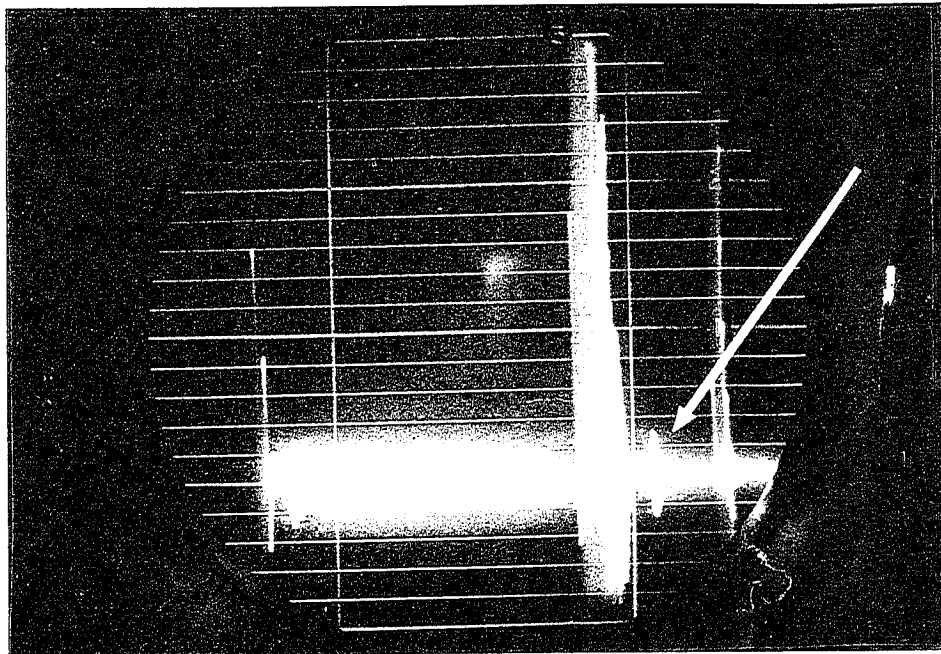


Fig. 5. Normal ultrasound reflection pattern from lucite cube; no modifying beam in operation, detection beam operating at a frequency of 5 Mc. Arrow points to internal reflection from structural inhomogeneity.

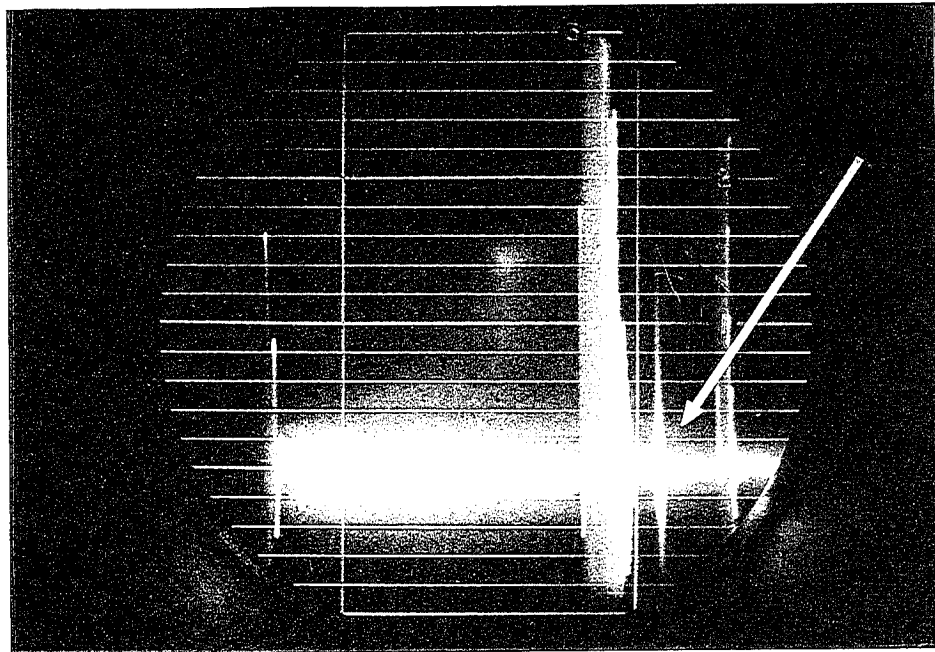


Fig. 6. Ultrasound reflection pattern following application of 3Mc modifying beam for 0.3 sec. Detecting beam operating at 5 Mc. Arrow points to transiently enhanced internal reflection from structural inhomogeneity. Same gain as Fig. 5.

at right angles to the detecting beam, and is positioned to irradiate the site of the existing internal inhomogeneity (which is just detectable by the ordinary ultrasonic method), is illustrated in Fig. 6. It is obvious from a comparison of Fig. 5 and Fig. 6 (the sensitivity settings are identical) that the amplitude of the reflection emanating from the region of the inhomogeneity is increased in Fig. 6. The normal reflection pattern recorded when the modifying beam was turned off showed no change from that illustrated in Fig. 5. When the focal point of the modifying beam was moved deeper into the lucite block a transiently induced reflection, possibly due to a previously undetectable inhomogeneity, was recorded as illustrated in Fig. 7*. Again when the modifying beam was turned off no such reflection was evident. (The reflectoscope gain was doubled for Fig. 7 as compared to that of Fig. 6.)

Fig. 8 shows the normal reflected pattern with the detecting beam set at a frequency of 2.25 Mc. This may be compared to Fig. 9 which shows the experimental pattern when a modifying beam of ultrasound of 3 Mc frequency is applied for 0.3 sec. Again, under these latter conditions either a structural feature previously undiscernible can be detected when a modifying ultrasound beam is applied, or a transient inhomogeneity was induced.

DISCUSSION

Although the above experimental results indicate the feasibility of the proposed method, it is important to realize that if full advantage is to be taken of the proposed system in order that it will have maximum usefulness as a biological and medical tool, major design and development must take place. The system now under construction is a precise laboratory model which is designed to give quantitative information on a variety of biological structures including that of brain. Since this first system is of the general purpose type designed to obtain working familiarity with this new method it necessarily must be more flexible than later models which would be designed for the examination of specific parts of the body or for specific organs. A minimum of three transducers (examining, receiving, heating) is being incorporated into the present instrumentation. Elliptic and hyperbolic acoustic reflectors which permit collection of energy scattered over a large solid angle to provide good azimuth resolution are included in the first transducer arrangement (Olofsson, 1963). Provision is also made that the examining assembly, with its associated acoustic reflectors and receiving transducer, can move in a wobbling motion in order to minimize the effects of ghost images. This transducer system with its supports is, of necessity, comparatively massive. The major system components are being designed not only for in vivo visualization of animal tissues but also for application to the human. It is, of course, necessary that boundaries be viewed at a variety of orientations if false echoes and artifacts are to be discriminated from identifying echoes. Since the first instrument is in the form of a general-purpose device so that a variety of transducer configurations and a range of possible uses can be demonstrated, it is necessary that the automatic scanning capability include three directions of linear motion and one rotational degree of freedom. The scanning must be accomplished in a relatively rapid fashion in order to obtain information for a complete scan cross section in a reasonable time interval and to provide information fast enough to simplify integration of signals. The requirement of this rectilinear movement of a relatively heavy array of acoustic instrumentation under accurately controlled conditions of motion has been met by an adaptation of a machine tool which provides accurate stationary position information ($\pm 0.001''$) in three orthogonal directions under code instructions from a paper tape. Linear motion potentiometers provide information on the details of the motion during movement. Analysis of more elaborate encoding methods for this type of machine have been made and are planned for application to future models. Since the entire sequence of electroacoustic events, which involves arrays

* The question of detecting a transiently induced inhomogeneity is of considerable interest but is not discussed here. Preliminary investigations on certain aspects of this question are discussed in a final report on contract Nonr-4800(00) (x) entitled Research on Ultrasonic Detection of Structural Inhomogeneities and Residual Stresses by F. J. Fry and E. Kelly.

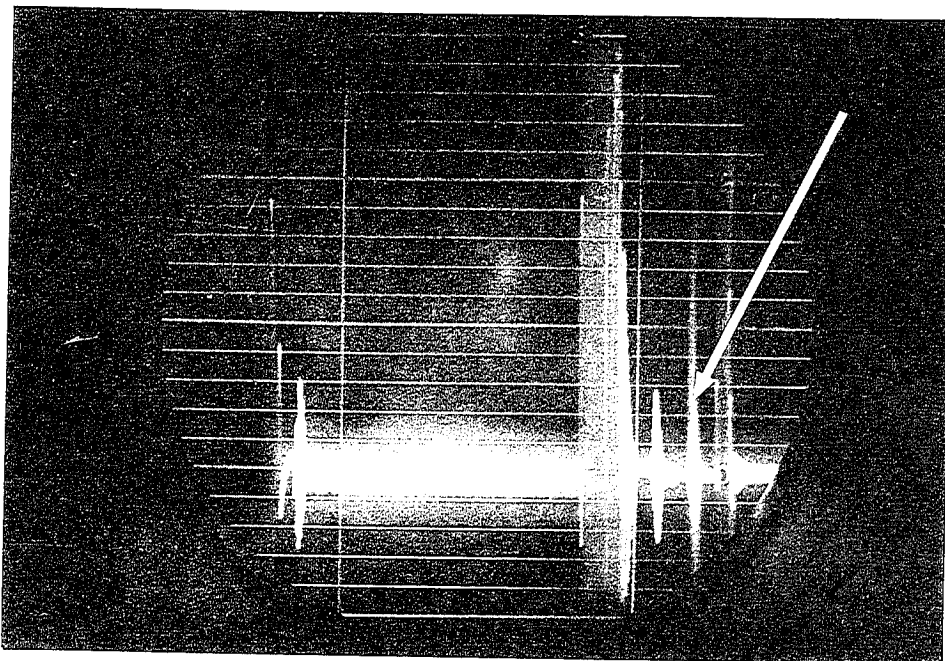


Fig. 7. Ultrasound reflection pattern following deeper application in lucite cube of 3 Mc modifying beam (as compared to Fig. 6). Detecting beam operating at 5 Mc. Gain increased by a factor of 2 as compared to Fig. 6. Arrow points to transiently induced internal reflection.

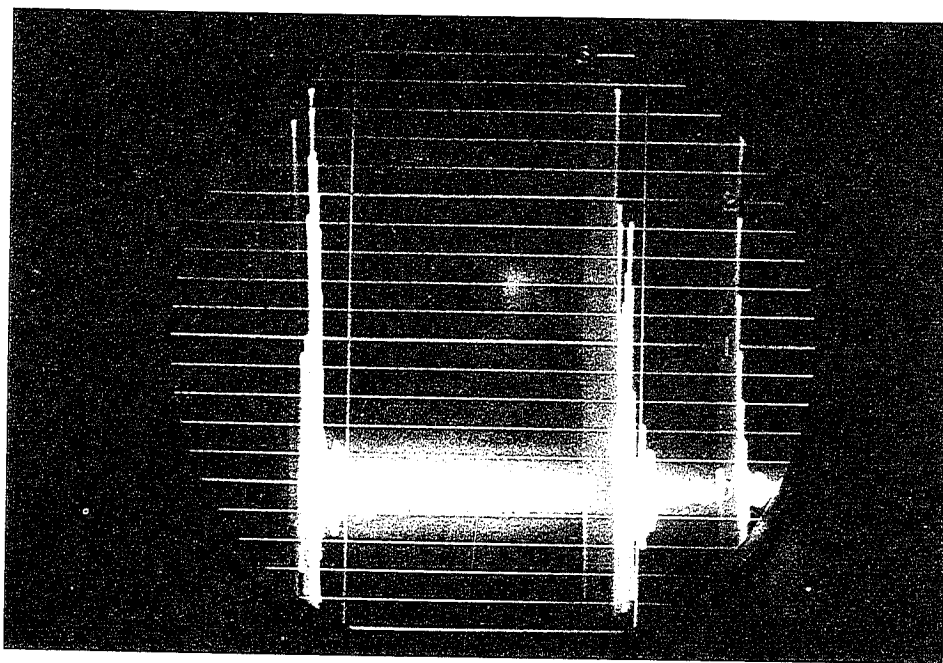


Fig. 8. Normal ultrasound reflection pattern from lucite cube; no modifying beam in operation, detecting beam operating at a frequency of 2.25 Mc.

of transducers as compared to present ultrasonic visualization systems, must be "synchronized" with a knowledge of the spatial positions of all of the transducers--both impedance changing and examining--and since other instruments are required to first generate and then accept and interpret the acoustic echoes received from a three-dimensional tissue target and then display the processed information in an appropriate visual manner with accurate reproduction of structural features, it is apparent that a rather sophisticated electronic system is required. Fig. 10 shows an overall view of the acoustic and electronic instrumentation at its present stage of development.

If ultrasonic visualization systems are to be developed to the full extent of their capability it is apparent that modern computers should be included in the instrumentation. A computer could perform a variety of complex tasks during tissue visualization including: preparation and control of complex scan paths for examining transducers, master control for recording and retrieval of data from such scans, processing the echo information at a computation rate sufficient to give an adequate display picture and correlation of the echoes from different transducer positions in order to select "true" physical features as distinguished from ghost echoes or noise. It is not intended to discuss these aspects in detail here but it may be of some interest to consider the general type of computer requirements to be used in a program of visualization of brain--a structure with a high density of features of interest. If the acoustic instrumentation discussed here provides a resolution of detail of the order of 1 mm, the possibility exists that of the order of 100 echoes per square centimeter projected area of brain tissue surface may be received from a single scanning transducer at a single observational position. Further, the use of multiple examining transducers should be considered in order to obtain maximum information. If only three such transducers were used, some 300 possible echoes per square centimeter of surface could be received. When equal resolution throughout a one-centimeter thickness of tissue is considered instead of a single plane, the number of echoes is of the order of 10,000. While it is obvious that not all of these echoes will be received always, nor will they necessarily all be collected prior to any processing by the computer, this type of estimation indicates that a computer of at least 4000 words of storage should be considered.

SUMMARY

A method of ultrasonic tissue visualization which is dependent upon differences in acoustic absorption coefficient values is described, and a discussion of some of the instrumentation under development for implementation of the new method is presented.

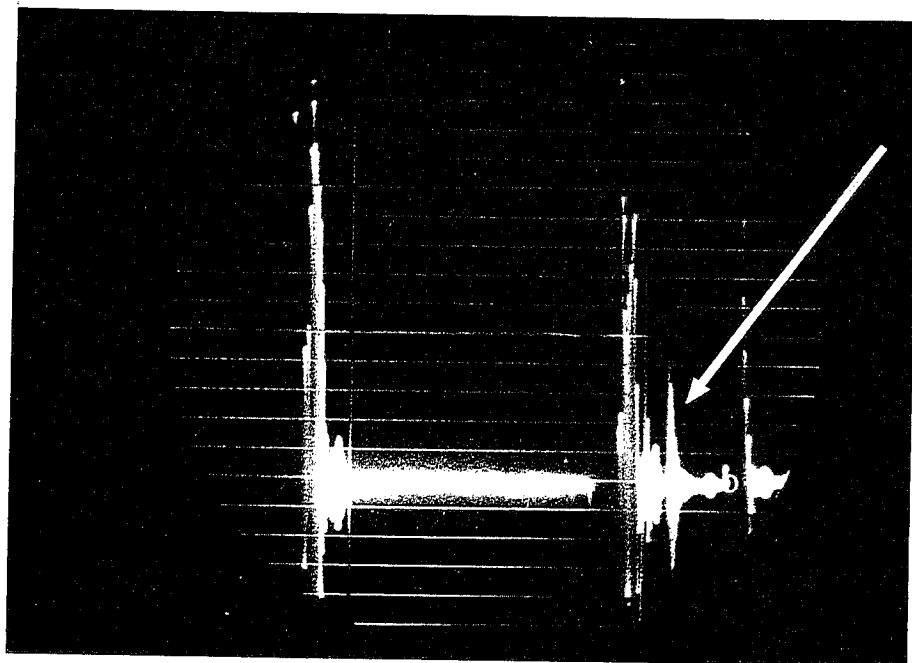


Fig. 9. Ultrasound reflection pattern following application of 3 Mc modifying beam for 0.3 sec. Detecting beam operating at 2.25 Mc. Arrow points to transiently induced internal reflection.

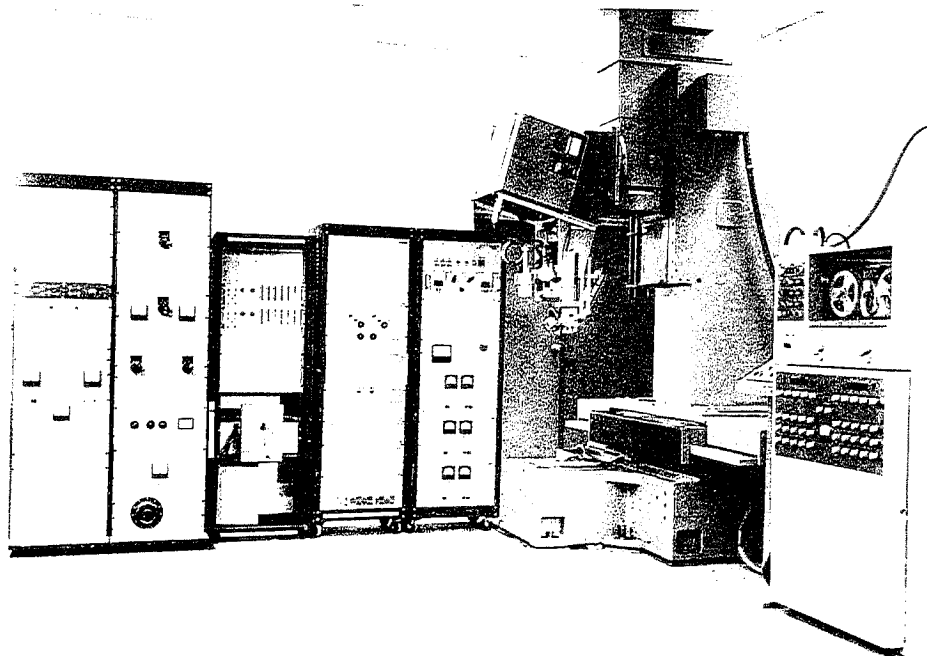


Fig. 10. Acoustic and electronic instrumentation used in new method of ultrasonic visualization.

REFERENCES

1. Auberger, M. and J. S. Rinehart (1961). Ultrasonic attenuation of longitudinal waves in plastics. J. Appl. Phys., 32:219.
2. Barnard, J. W., W. J. Fry, F. J. Fry and R. F. Krumins (1955). Effects of high intensity ultrasound on the central nervous system of the cat. J. Comp. Neurol., 103:459.
3. Baum, G. and I. A. Greenwood (1965). Current status of ophthalmic ultrasonography in Ultrasonic Energy--Biological Investigations and Medical Applications. E. Kelly, ed., University of Illinois Press, Urbana, Illinois.
4. Breazeale, M. A. and D. O. Thompson (1963). Finite-amplitude ultrasonic waves in aluminum. Appl. Phys. Ltrs., 3:77.
5. Carstensen, E. L., K. Li and H. P. Schwan (1953). Determination of the acoustic properties of blood and its components. J. Acoust. Soc. Am., 25:286.
6. Dunn, F. (1965). Ultrasonic absorption by biological materials in Ultrasonic Energy--Biological Investigations and Medical Applications. E. Kelly, ed., University of Illinois Press, Urbana, Illinois.
7. Espinola, R. P. and P. C. Waterman (1958). Ultrasonic interferometer for the measurement of the temperature dependence of elastic constants. J. Appl. Phys., 29:718.
8. Fry, W. J. (1962). Present and future applications of ultrasonics in biomedicine. Proc. IRE, Vol. 50:1393.
9. Fry, W. J. (1965). New approaches to the study and modification of biological systems by ultrasound in Ultrasonic Energy--Biological Investigations and Medical Applications. E. Kelly, ed., University of Illinois Press, Urbana, Ill.
10. Fry, W. J. and F. Dunn (1962). Ultrasound: analysis and experimental methods in biological research in Physical Techniques in Biological Research. W. L. Nastuk, ed., Academic Press, New York, Vol. IV, Chapt. 6.
11. Fry, W. J. and F. J. Fry (1963). Ultrasonic visualization of soft tissue structure based on gradients in absorption characteristics. J. Acoust. Soc. Am., 35:1788.
12. Margenau, H. and G. M. Murphy (1943). The Mathematics of Physics and Chemistry. D. Van Nostrand Co., Inc., New York.
13. Mason, W. P. and H. J. McSkimin (1947). Attenuation and scattering of high frequency sound waves in metals and glasses. J. Acoust. Soc. Am., 32:464.
14. Olofsson, S. (1963). An ultrasonic optical mirror system. Acustica, 13:361.
15. Papadakis, E. P. (1960). Ultrasonic attenuation in S.A.E. 3140 and 4150 steel. J. Appl. Phys., 32:1628.
16. Papadakis, E. P. and E. L. Reed (1961). Ultrasonic detection of changes in the elastic properties of a 70-30 iron-nickel alloy upon heat treatment. J. Appl. Phys., 32:686.
17. Roderick, H. L. and J. Truell (1952). The measurement of ultrasonic attenuation in solids by the pulse technique and some results in steel. J. Appl. Phys., 23:267.