Focusing system
for an ultrasonic transducer

by

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INTRODUCTION

Diagnostic ultrasound is one technique of visualizing organs of the body by recording the reflected part of the transmitted sound energy from the interfaces of different tissues; a radar system detects the interface of air and metal. A piezoelectric crystal transforms short pulses of electrical energy to mechanical vibrations, sound energy, that penetrate the body. The reflections from each interface of different tissues are transformed to electrical energy by the same or another piezoelectric crystal. The term ultrasonic transducer refers to the piezoelectric crystal of the transducer that operates at a frequency higher than 20 KHz.

The ultrasonic image quality of an object depends on the properties of the ultrasonic transducer. A transducer should have three major optimum properties: sensitivity, axial resolution, and lateral resolution. Transducer sensitivity is related to the detection of small tissue interfaces at certain depths in the body. The axial and lateral resolution are the detection of the distance between the interfaces in the axial and normal planes respectively, as shown in Figures 1 and 2.

Background

Most research revolves around achieving the optimum properties of the transducer. As a result, several transducers with varying geometrical shapes and materials have been developed.

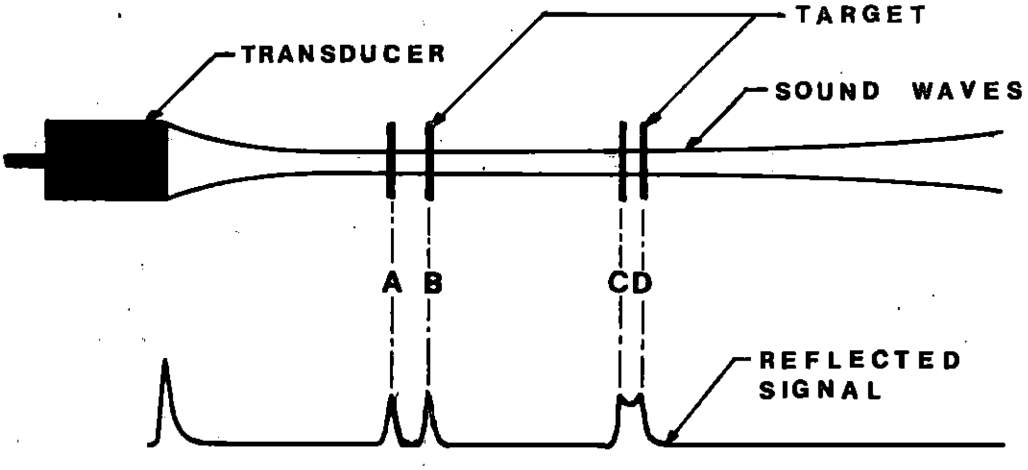

Figure 1. Principle of axial resolution.

The axial resolution permits detection of A and B as separate targets, but targets C and D are not resolved due to insufficient distance between the two targets.

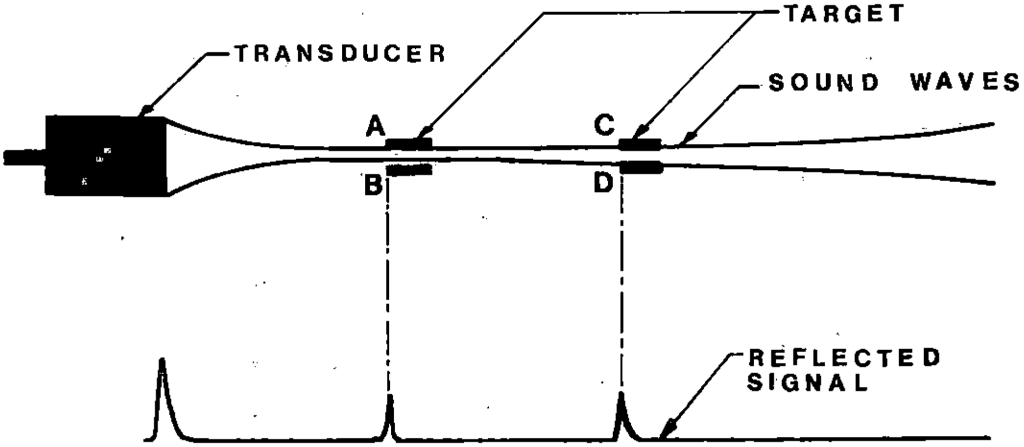

Figure 2. Principle of lateral resolution.

The lateral resolution permits detection of A and B separately as the transducer moves downward. Targets C and D are not resolved independently since the beam width of the transducer is greater than the separation between the two targets.

The transducer sensitivity depends on the nature of the piezoelectric crystal, the damping of the transducer, focusing power, and the distance and nature of the reflector. Several types of piezoelectric crystals are available for electrical and mechanical energy transformation; the ceramic crystals have good sensitivities (9). An under-damped transducer, high Q-factor, enhances the sensitivity by having more available power to be transmitted (9). The focused transducer has more available power for reflection in the focal region, hence, improving the sensitivity.

The axial resolution depends on the frequency and the damping of the transducer. An increase in the frequency of the pulse increases both the axial resolution and the attenuation of the ultrasonic signal by the tissue of the body. Tissue absorption of sound energy is proportional to the pulse frequency (9). A compromise between the axial resolution and penetration depth determines the frequency. Also, axial resolution can be improved by damping the system, low Q-factor, to reduce the pulse duration at the expense of the sensitivity (1), as shown in Figure 3.

Lateral resolution depends on the piezoelectric geometry, frequency, and focusing. A plane transducer can have one of two types of crystals, a circular crystal, or a disc-ring crystal. A disc crystal has a uniform lateral resolution in the near field and a divergent sound beam in the far field (10). The lateral resolution can be improved by

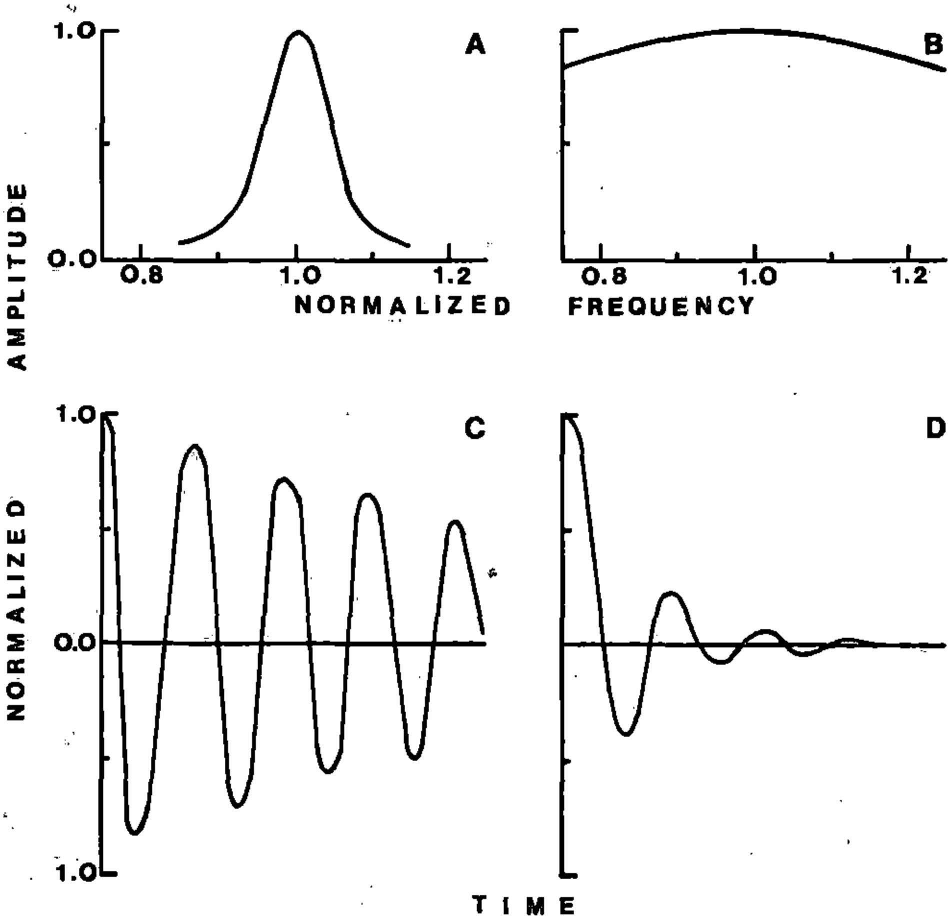

Figure 3. Effect of Q-factor on transducer response.

(A),(C)-high Q-factor transducer has under-damped time response; the transducer is most sensitive at the natural frequency. (B),(D)-low Q-factor transducer has over-damped time response; axial resolution is enhanced with decreasing signal duration. (From Wells (9)) increasing the frequency or the piezoelectric crystal diameter; both decrease the rate of divergence (9). The disc-ring transducer gives a uniform lateral resolution at certain distances from the transducer by driving the inner disc crystal and the outer ring crystal at different voltage ratios (4).

The focused transducer has several configurations: curved ceramic, simple lens, and two lens system. The curved ceramic uses a concave hemisphere piezoelectric crystal having a non-uniform beam width and a better lateral resolution than the plane transducer due to reduced beam width in the focal region (6, 11). The application of lenses gives the same resolution capability as the curved ceramics but at a reduced cost. The simple solid lens transducer uses a concave lens because the velocity of sound waves is greater in solids than in liquid medium or the tissue (8). The two lens system consists of a solid plano-concave lens and a liquid lens that modifies the focal distance (7). The lens system transducer reduces the sensitivity because of increased absorption losses by the lenses and increased internal reflections at the interface of lenses. In order to minimize the absorption losses, zone lenses can be used. The zone lens, with stepped thickness of half wavelengths, decreases the losses by reducing the wave interference in the lens (8). A quarter wavelength matching layer can be used to minimize the internal reflections (1, 8).

Regardless of which type of transducer is used, each has a flat face at the interface with the object being visualized.

Objective of Research

The objective of this research is to develop a transducer that can be easily moved along the curved shape of the body, and has an improved lateral resolution. A convex transducer face is used to enhance the mobility of the transducer along the body and a focusing system is implemented to improve the lateral resolution. Several modified transducers can be mounted on a scanning bar to reduce the examination time and enhance the mobility as shown in Figure 4.

The basic properties of the sensitivity and axial resolution are discussed for future modifications of ultrasonic transducers; however, they are not the focal point of this research. In developing this transducer, the sensitivity and axial resolution modifications are implemented to obtain an acceptable response.

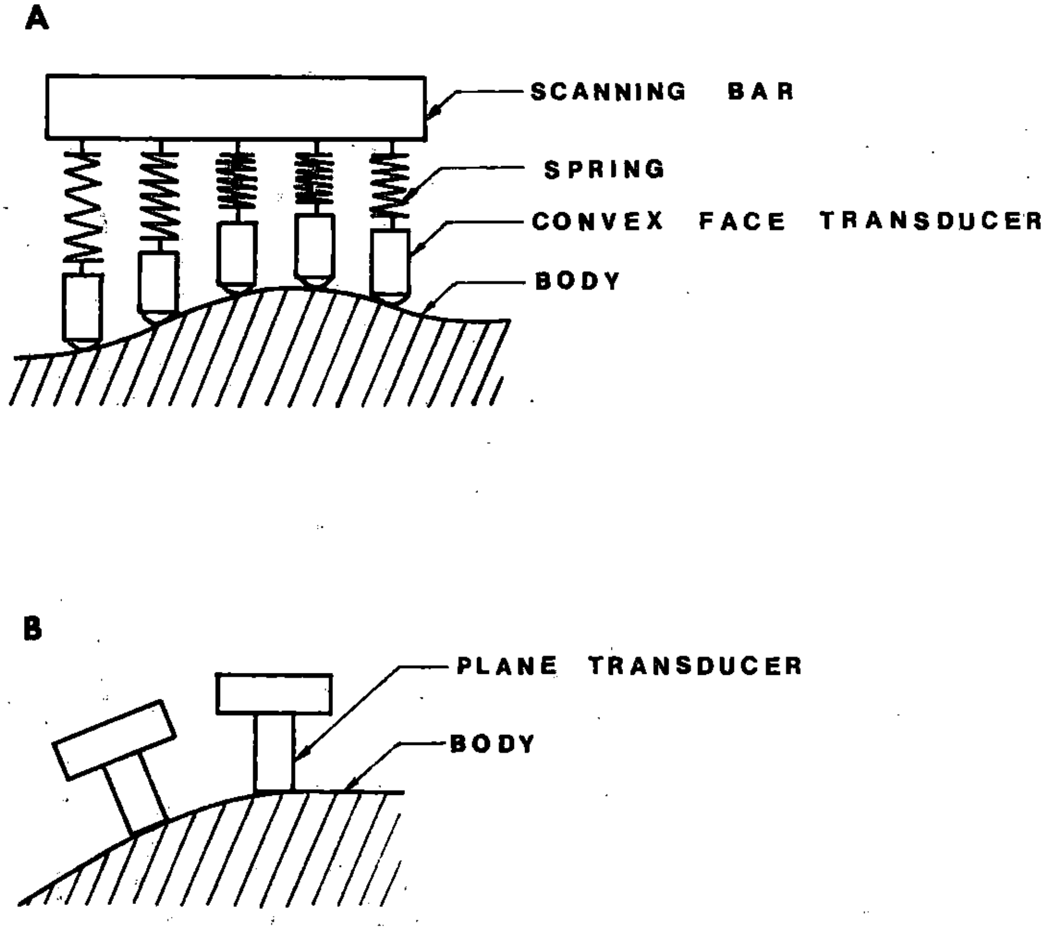

Figure 4. Schematic of scanning bar showing mobility of transducer.

(A)- Scanning bar has the advantage of reducing examination time; convex face transducers are mounted on the bar by a spring system, enhancing mobility of the transducers along the curved shape of the body.

(B)- The plane face transducer is more difficult to move along the body since it needs tilting to ensure good contact with the body.

DESIGN

The transducer design is worked with an emphasis on a convex surface of the transducer for the enhancement of contact with and movement along the body. The effect of the convex surface is divergence of the sound energy, so a concave lens is used to give an enhanced lateral resolution by its convergence effect, as shown in Figure 5. A damping medium is placed against the piezoelectric crystal to improve axial resolution. Sensitivity is improved by tuning the transducer with a shunt inductor and using a matching layer to reduce multiple internal reflections. The design process includes improving sensitivity, axial resolution, and lateral resolution for a transducer that acts as a transmitter and a receiver.

Sensitivity and Axial Resolution

Since the transducer acts as a transmitter and receiver, a lead zirconate titanate, PZT -5, crystal is used for its good sensitivity performance compared to other piezoelectric crystal sensitivities (9). The improvement of sensitivity and axial resolution is made by mechanical and electrical impedance matching. The mechanical impedance matching requires a backing material for the piezoelectric crystal and a matching layer between the transducer and the load. The backing material acts as a damping medium of the crystal vibrations. Air-backing medium reflects most of the energy due to its low impedance compared with the crystal; as a result, maximum power is reflected to the crystal and thus enhancing the sensitivity. One disadvantage of using air-backing

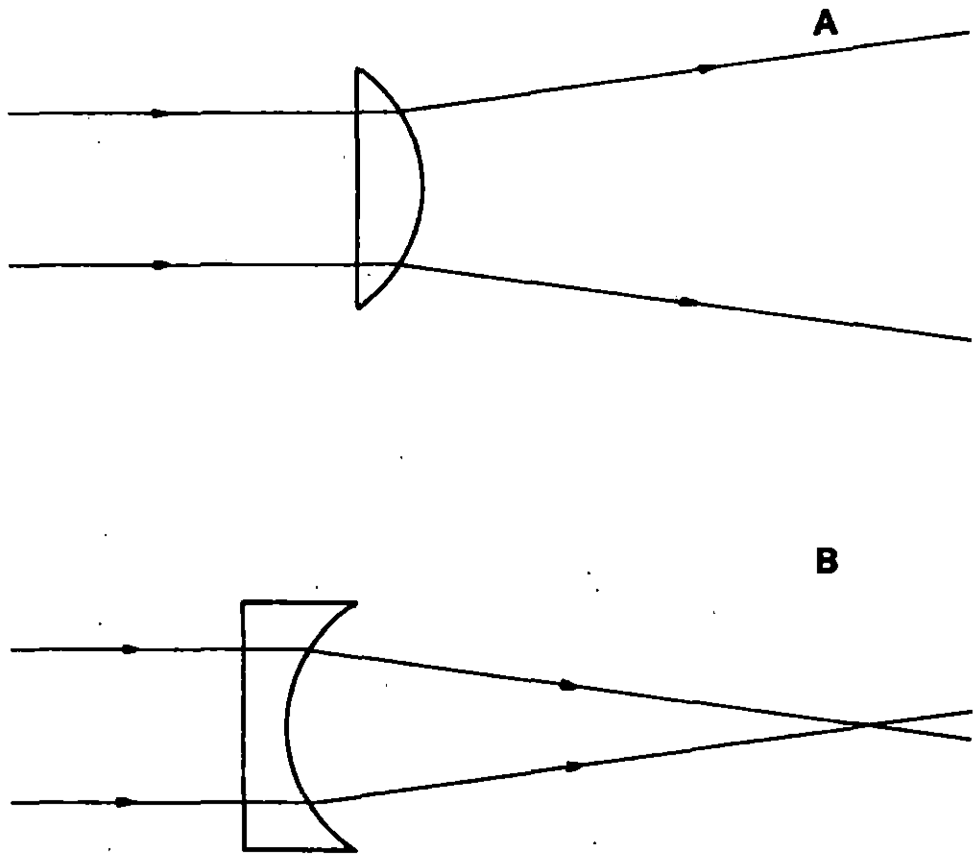

Figure 5. Effect of solid lens in liquid medium.

Velocity of sound is greater in a solid medium than in a liquid medium or tissue. As a result, the convex lens A is a diverging lens and the concave lens B is a converging lens.

medium is the narrow bandwidth, high Q-factor, of the vibrations due to under-damped effect of air medium (1). A wide bandwidth is required for good axial resolution; this is achieved by a damping material that consists of tungsten powder and epoxy mixture (1, 9). The ratio of the tungsten powder and epoxy mixture determines the degree of the required damping that gives a compromise between the sensitivity and axial resolution. A mechanical impedance transformer between the transducer and the load improves the sensitivity by reducing the reflected losses at the interface of the transducer and the load. The mechanical impedance transformer has an intermediate impedance between the two interface impedances and a thickness of a quarter wavelength (1, 8). The electrical impedance matching is done by a tuning shunt inductor, which resonates with the capacitance of the crystal and the echo-scope, to give a maximum electrical efficiency that improves the sensitivity (1, 9). The sensitivity is also improved by the focusing system that is used to get a good lateral resolution.

Lateral Resolution

The convex outer surface of the transducer is used as a part of the focusing system. The lateral resolution is enhanced by three lenses: two solid lenses and one liquid lens as shown in Figure 6.

Plano-concave solid lens

A plane surface of the lens has been chosen because it minimizes the reflected energy loss by the direct contact with the piezoelectric crystal (6). The concave surface is the main source of beam convergence.

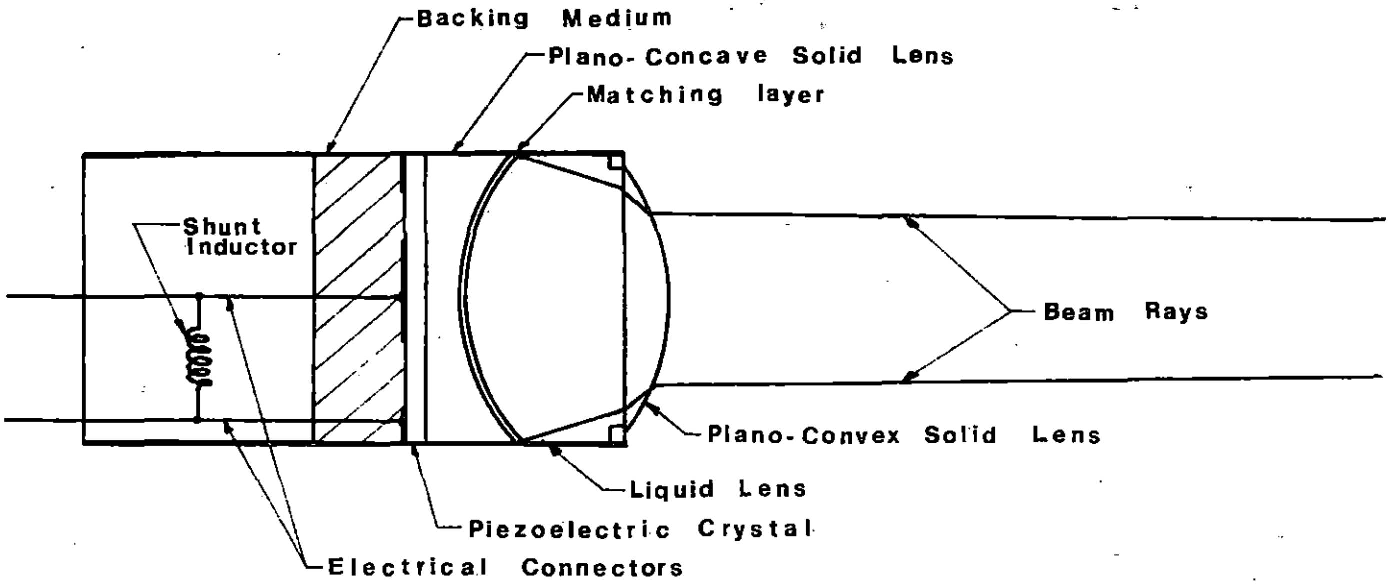

Figure 6. Transducer construction and its effects on beam convergence.

The lens should have minimum thickness to reduce the absorption loss of the solid material. A lens that has a multiple half wavelength thickness at the axial position gives the least loss (8, 10); the increased thickness at the edges of the lens reduces the side lobes of the beam (3, 7).

Plano-convex solid lens

The convex shape is required for an easy movement along the body. A plane surface with the minimum lens thickness is implemented to further reduce losses (7). This lens diverges the sound beam; however, the overall effect of the three lenses is designed to give a final convergence.

Plano-convex liquid lens

The liquid lens takes the plano-convex shape because the gap between the two solid lenses is filled with liquid. The required refractive index and an easy fabrication of the whole system are the results of chosing the liquid lens.

Primary Steps

The purpose of the design primary steps is to evaluate different options that a designer will encounter. Each design has different limitations and dimensions that depend on the material of the transducer.

Limitations

The basic limitations are the nature of the lenses and the dimensions of the transducer. Lens requirements are

1. low absorption coefficient,
2. high refractive indexes, critical refractive angles,
3. matched acoustical impedances,
4. dimensions for easy fabrication.

Transducer dimensions determine the spot size at the focus (see below) and the degree of convergence.

Materials

Several transducers were fabricated with the combination of the following materials:

1. piezoelectric crystals, 2.2 MHz lead zirconate titanate, PZT-4 and PZT-5
2. backing medium
    a. air
    b. tungsten powder and epoxy mixture
        1. 2:1 weight ratio
        2. 1:1 weight ratio

3. piezoelectric crystal dimension, 6.35mm and 12.7mm in diameter.
4. lenses
    a. aluminum and mercury
    b. Plexiglas and water

    All lenses are hemispherical in shape; dimensions of the lenses vary with each focusing system.
5. matching layer, a quarter wavelength of paraffin wax is used at the interface of Plexiglas and water lenses.

Focusing system

The first design of the focusing system included two aluminum lenses and a mercury lens. These materials were chosen because of the high refractive indexes, low absorption losses, and small impedance mismatch between aluminum and mercury. This design was abandoned for two reasons: the hazardous effects of mercury and the low sensitivity due to great impedances mismatch of aluminum and tissue. An impedance matching layer for the aluminum-tissue interface requires a skillful fabrication to be implemented because a solid layer is required.

The final design includes two Plexiglas lenses and a water lens. This combination was chosen because of the Plexiglas-water refractive index, the low absorption loss by water, and the good overall transducer dimensions. Each lens thickness was minimized to reduce the absorption loss of Plexiglas. A paraffin wax matching layer was used to reduce the loss that was due to impedance mismatch of Plexiglas and water.

Theoretical Design

Governing equations

The following equations determine the beam pathway, the spot size, and the effect of the matching layer:

1. Snell's Law: $\frac{\sin\theta i}{ci} = \frac{\sin\theta t}{ct}$ Eq. 1

   where:

   θ = angle of incidence or refraction
   c = velocity of sound waves in medium

2. Spot size at the focus: $z = 2.44\,\frac{\lambda f}{d}$ (3, 10) Eq. 2

   where:

   λ = wavelength of medium
   f = focal distance
   d = diameter of the lens

3. Power transmission coefficient, α_t, for three lossless media at normal incidence(9):

$$\alpha_t = \frac{4Z_3Z_1}{(Z_3 + Z_1)^2 \cos^2 K_2 l_2 + \left(Z_2 + \frac{Z_3Z_1}{Z_2}\right)^2 \sin^2 k_2 l_2} \qquad \text{Eq. 3}$$

where:

$k_2 = \frac{2\pi}{\lambda}$, wavelength constant of second medium
λ = wavelength
Z_i = acoustical impedance of medium
l_2 = thickness of second medium

The power transmission coefficient is the ratio of the transmitted to the incident signal power. This equation is an approximation since the Plexiglas lens and the paraffin layer have absorption losses.

Calculations

The following theoretical calculations predict the effect of the paraffin wax matching layer and the expected beam pathway. All values are taken from Table 1.

For Plexiglas and water lenses with paraffin wax matching layer:

Z_1 = impedance of Plexiglas

Z_2 = impedance of paraffin wax

Z_3 = impedance of water

Power transmission coefficient

Case 1 Without a matching layer, the solution of Eq. 3 is:

$l_2 = 0$

$$\alpha_t = \frac{4Z_3Z_1}{(Z_3+Z_1)^2} = 0.869$$

Case 2 With a matching layer, the solution of Eq. 3 is:

for a quarter wavelength , $l_2 = \frac{\lambda}{4}$, thickness of paraffin wax matching layer

$$\alpha_t = \frac{4Z_3Z_1}{\left(Z_2+\frac{Z_3Z_1}{Z_2}\right)^2} = 0.992$$

matching layer thickness:

$$\lambda = \frac{\text{velocity}}{\text{frequency}} = \frac{2X10^6}{2.2X10^6}\ \frac{\text{mm/sec}}{\text{Hz}} = 0.9091\ \text{mm}$$

$$l_2 = \frac{\lambda}{4} = 0.227\ \text{mm}$$

The paraffin wax matching layer of quarter wavelength thickness improves the ratio by 14.2 per cent.

Table 1: Velocity of sound and acoustic impedance of materials.

(From Krautkrämer and Krautkrämer (5))

Material	Velocity of sound m/sec	Acoustic impedance 10^6 kg/m^2-sec
Aluminum	6320	17.0
Mercury	1450	20.0
Plexiglas	2730	3.2
Water	1483	1.5
Paraffin wax	2200	2.0

Beam pathway

Since the matching layer has very small thickness, it has a negligible effect on changing the beam pathway; as a result, the analysis is based on Figure 7 with the application of Eq. 1 and 2. The calculation is based on angles that correspond to beam pathway at each interface. The tissue is replaced by water since testing the transducer is easily done in a water tank; both have almost the same velocity of sound, but tissue has a higher absorption coefficient than water.

$$R\sin\theta_1 = \sin\theta_2 \qquad (1a)$$

$$R\sin\theta_4 = \sin\theta_3 \qquad (1b)$$

$$R\sin\theta_5 = \sin\theta_6 \qquad (1c)$$

where $R = C_{water}/C_{Plexiglas}$

$$\theta_1 - \theta_2 = \theta_3 \qquad (4)$$

$$\theta_4 + \theta_6 - \theta = \theta_5 \qquad (5)$$

$$Z = 2.44\,\frac{\lambda f}{d} \qquad (2a)$$

$$\tan\theta = \frac{d/2}{f} = \frac{do/2}{fo} \qquad (6)$$

condition: $d > do > Z$

where:

d = crystal diameter

f = effective focal length

do = beam diameter at last interface

fo = distance from last interface to focal region

Before solving the above equations, the best combination of crystal diameter, effective focal length, spot size, and critical angle of refraction must be determined. After finding the best combination that

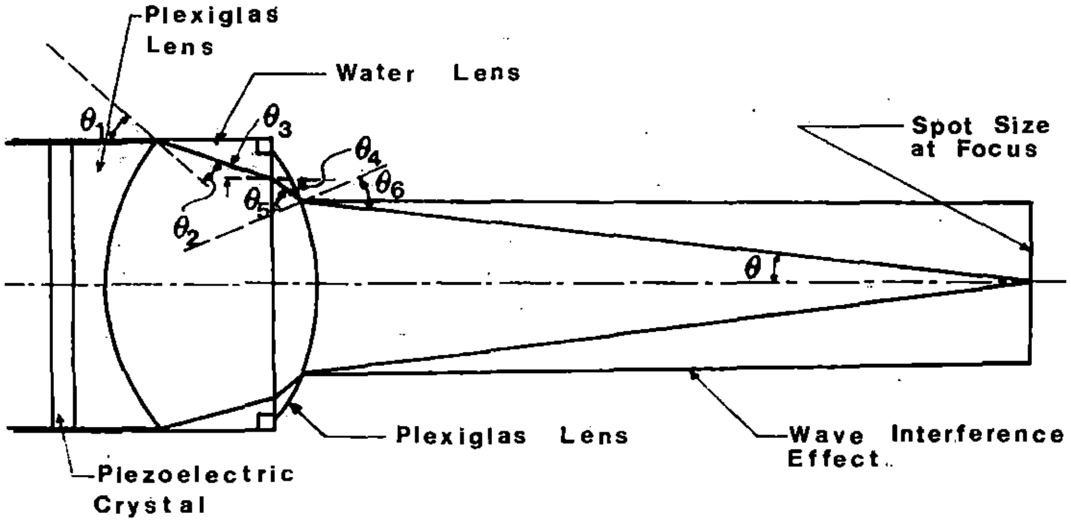

Figure 7. Theoretical ray pathway.

The ray pathway is found by considering the angles at each interface and the affect of wave interference that results in a finite spot size.

satisfies condition 7, the beam pathway is found easily from equations 1a, 1b, 1c with the restriction on the value of $(\theta_6 - \theta)$ to be less than the critical refractive angle, 33.37°. The solution of equations 2a and 6 with condition 7 is:

spot size = Z = 7.1 mm

beam diameter at last interface = do = 7.2 mm

distance from last interface to focal region = fo = 30.9 mm

crystal diameter = d = 12.7 mm

effective focal length = f = 54.5 mm

$\theta = 6.65^\circ$.

The solution of equations 1, 4, and 5, with $(\theta_6 - \theta) < 33.37^\circ$, requires the assumption of one angle because there are five equations and six unknowns. The assumed angle is manipulated to satisfy the critical angle of refraction and to give good lenses dimensions that can be easily manufactured. The assumption is made to give θ_6 a value of 28.86°. The values of the other angles are easily found to be:

$\theta_5 = 61.32^\circ$

$\theta_4 = 39.12^\circ$

$\theta_3 = 20.30^\circ$

$\theta_1 = 41.82^\circ$

$\theta_2 = 21.51^\circ$

The lenses dimensions can now be easily calculated from the corresponding angles. Radius of curvature of each lens is equal to 9.525 mm. The thickness of the plano-concave lens at the axial position is made equal to approximately one wavelength, 1.24 mm, because the reflected waves at each interface of the lens are in-phase. The thickness of the Plexiglas plano-convex lens at the axial position is made equal to 1.78 mm; this value requires a thickness of 7.5 mm for the water lens.

Experimental Procedure

Several transducers were tested. The alumimmm -mercury transducer had a very low sensitivity due to internal reflection losses. The PZT-5 crystal had a better sensitivity than the PZT-4 crystal. The transducer that has the quarter inch piezoelectric crystal gave a divergent beam due to diffraction. The damped transducer has good sensitivity and axial resolution with the one to one weight ratio of tungsten powder and epoxy mixture. The final design gives satisfactory results with the following material and dimensions:

1. backing medium, tungsten and epoxy mixture, one to one weight ratio
2. piezoelectric crystal, 2.2 MHz lead zirconate titanate, PZT-5
3. lenses
    a. plano-concave and plano-convex Plexiglas lenses
    b. plano-convex water lens
4. matching layer, paraffin wax
5. electrical tuning, 3.2uH shunt inductor
6. dimensions
    a. radius of curvature of each lens = 9.525 mm
    b. thickness at axial position:
        1. plano-concave lens, t = 1.24mm

    2. plano-convex lens, t = 1.78 mm
    3. water lens, t = 7.5 mm

- c. diameter of piezoelectric crystal = 12.7 mm
- d. diameter of plano-concave lens = 12.7 mm
- e. diameter of plano-convex Plexiglas lens = 11.1 mm
- f. thickness of matching layer = 0.227 mm (theoretical).

The transducer was fabricated with satisfactory dimensions, 0.01 mm accuracy. The thickness of the matching layer was prepared by forcing a sphere against the paraffin wax and plano-concave lens until the internal reflections reached a minimum amplitude.

METHODS AND RESULTS

Sensitivity

There is not a standard and defined method for measuring the sensitivity. The sensitivity of the transducer is good enough to detect 0.5mm copper wire at 4.5 cm from the transducer with 200mv peak-to-peak voltage.

Axial Resolution

Figure 8 shows the set up used to measure axial resolution. The time at which the signal decays to 10%, -20db, determines the axial resolution. Figure 9 shows the reflected signal from a flat aluminum plate at 2.5 cm from the transducer. The axial resolution of the transducer is equal to 6 mm. Figures 10 and 11 show the reflected signals from the palm of the hand at the base of the thumb.

Lateral Resolution

One standard method for measuring the lateral resolution is the beam width at which the reflected signal amplitude is equal to half the maximum response amplitude, -6db amplitude. Figure 8 shows the method to measure the beam width. The reflecting target is a cylindrical copper wire with a 0.5 mm diameter. The wire is moved horizontally to find the one half amplitude response for each distance from the transducer, vertical distance. Figure 12 shows that the lateral resolution is uniform between 2 cm and 3.5 cm with a beam width equal to 2.8 mm; the penetration depth, -6db maximum amplitude, is 6.0 cm.

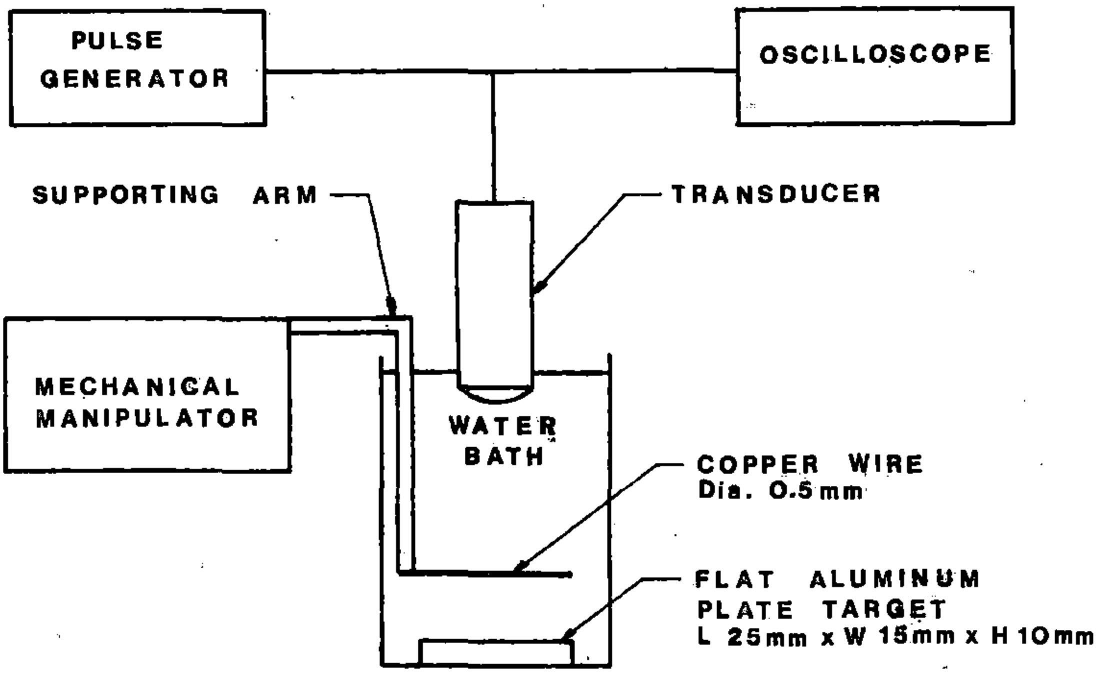

Figure 8. Method of measuring axial and lateral resolution.

Aluminum flat target is used to measure axial resolution from the reflected signal.
The copper wire is used to map the beam width and lateral resolution of -6db amplitude. The wire movement is controlled by a mechanical manipulator with a scale of 0.1mm accuracy.

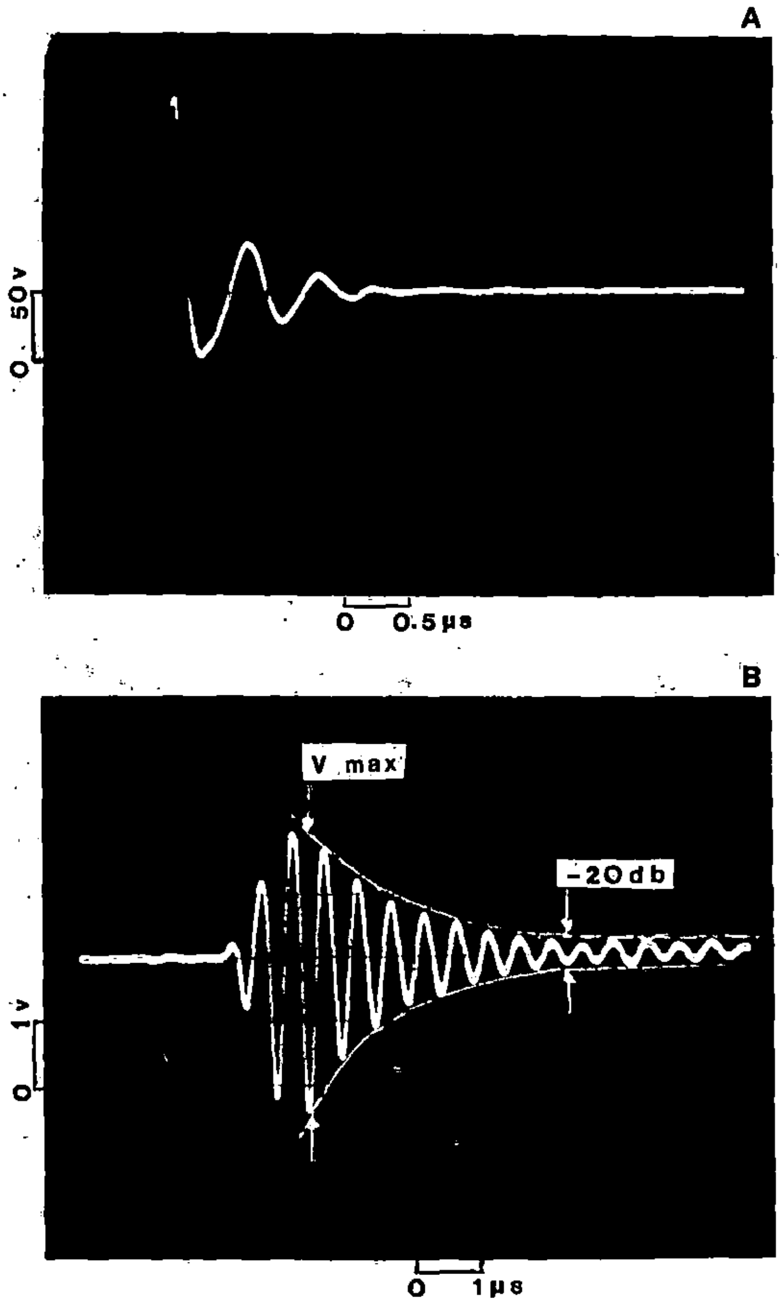

Figure 9. Transmitted and reflected signals.

(A) The transmitted signal, over-damped excitation pulse.
(B) The reflected signal from a flat target determines the axial resolution at the -20db amplitude; the equivalent distance is based on the average velocity of sound in tissue, 1500m/sec, and the corresponding time.

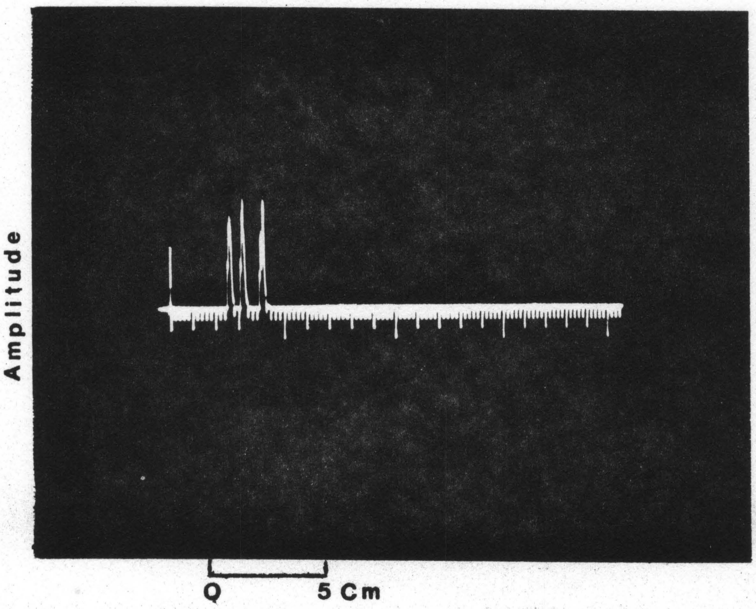

Figure 10. Echo-scope display of the reflected signal from palm of the hand at the base of the thumb. The reflected signal shows three layers of tissue separated by 6mm and 8mm.

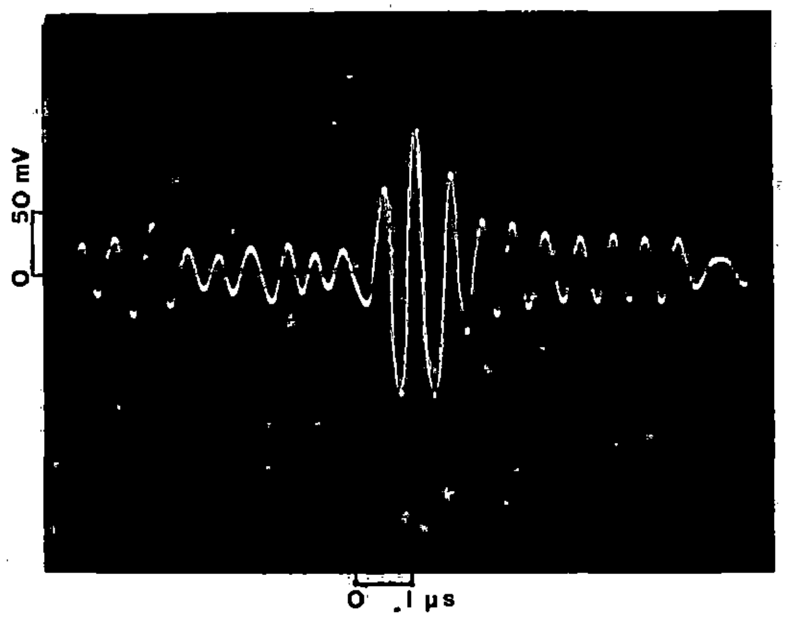

Figure 11. The actual shape of the second reflected signal from the palm of the hand. The signal duration has a corresponding 6 mm axial resolution.

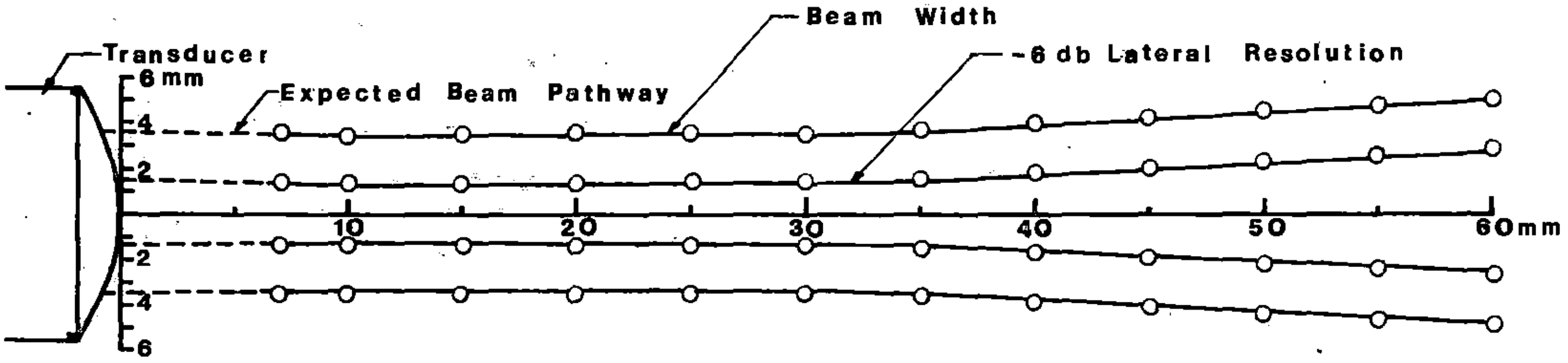

Figure 12. Beam width and lateral resolution.

Mapping the beam width from the amplitudes of the reflected signals is shown for distances greater than 7mm. Reflected signals from the wire, for distance less than 7mm, are superimposed on the reflected signal from the last interface of the convex lens; axial resolution is 6mm.

DISCUSSION AND RECOMMENDATIONS

The results show that a transducer with a convex outside surface has satisfactory properties except the penetration depth. The penetration depth can be increased at the expense of the lateral resolution. Increased penetration depth requires weak focusing that gives variable beam widths with increasing distances from the transducer. Also the beam width gets larger in the focal region with weak focusing. A transducer with larger dimensions gives good penetration depth and lateral resolution. A simple lens transducer with 2.8cm in crystal diameter and 25cm in lens curvature has 8mm beam width at 15cm of penetration depth (2). An axial resolution of 3mm could be achieved by the appropriate damping material (3). When using the plane transducer, 2.2MHz-19.05mm in diameter probe of Harisonic Medical, the reflected signal from the copper wire was 800 mv peak-to-peak voltage. However, the designed transducer had a sensitivity that permitted measurements of axial and lateral resolution. The reflected signal from the palm of the hand indicates an agreement with axial resolution measurement and a good sensitivity to detect tissue layers.

Theoretically, the final design could be improved by:

1) larger dimensions to achieve better penetration depth,

2) different damping material to achieve better axial resolution,

3) other combinations of lenses materials to achieve better sensitivity.

The final design achieves its purpose by having a 2.8mm lateral resolution and a convex surface to enhance movement along the skin above the spine; however, a greater penetration distance is required for imaging the whole spinal area. The design strategy developed is workable for any transducer having the lens system described in this thesis as can be seen from the agreement of the theoretical calculations and the beam width measurement.

REFERENCES

1. Kossoff, G. 1966. The effects of backing and matching on the performance of piezoelectric ceramic transducers. I.E.E.E. Trans. Sonics Ultrason. SU-13:20-30.

2. Kossoff, G., Robinson, D. E., Liu, C. N., and Garrett, W. J. 1964. Design criteria for ultrasonic visualisation systems. Ultrasonics 2:29-38.

3. Kossoff, G., Robinson, D. E., and Garrett, W. J. 1965. The C.A.L. abdominal echoscope. C.A.L. (Commonwealth Acoustical Laboratory, Department of Health), Miller's Point, Sydney, Australia, Report No. 31.

4. Kossoff, G. 1971. A transducer with uniform intensity distribution. Ultrasonics 9:196-200.

5. Krautkrämer, J., Krautkrämer, H. 1969. Ultrasonic Testing of Materials. Springer-Verlag, New York Inc., New York. Translation of the second revised German edition.

6. O'Neil, H. T. 1949. Theory of focusing radiators. J. Acoust. Soc. Am. 21:516-526.

7. Sette, D. 1949. Ultrasonic lenses of plastic materials. J. Acoust. Soc. Am. 21:375-381.

8. Tarnóczy, T. 1965. Sound focusing lenses and waveguides. Ultrasonics 3:115-127.

9. Wells, P.N.T. 1969. Physical Principles of Ultrasonic Diagnosis. Academic Press, London.

10. Wells, P.N.T. 1977. Biomedical Ultrasonics. Academic Press, London.

11. Willard, G. W. 1949. Focusing ultrasonic radiators. J. Acoust. Soc. Am. 21:360-375.

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