THE ROLE OF ULTRASONOGRAPHY IN THE DIAGNOSIS OF ENDOCRINE EXOPHTHAMOS

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M.D. Degree in Ophthalmology

BY

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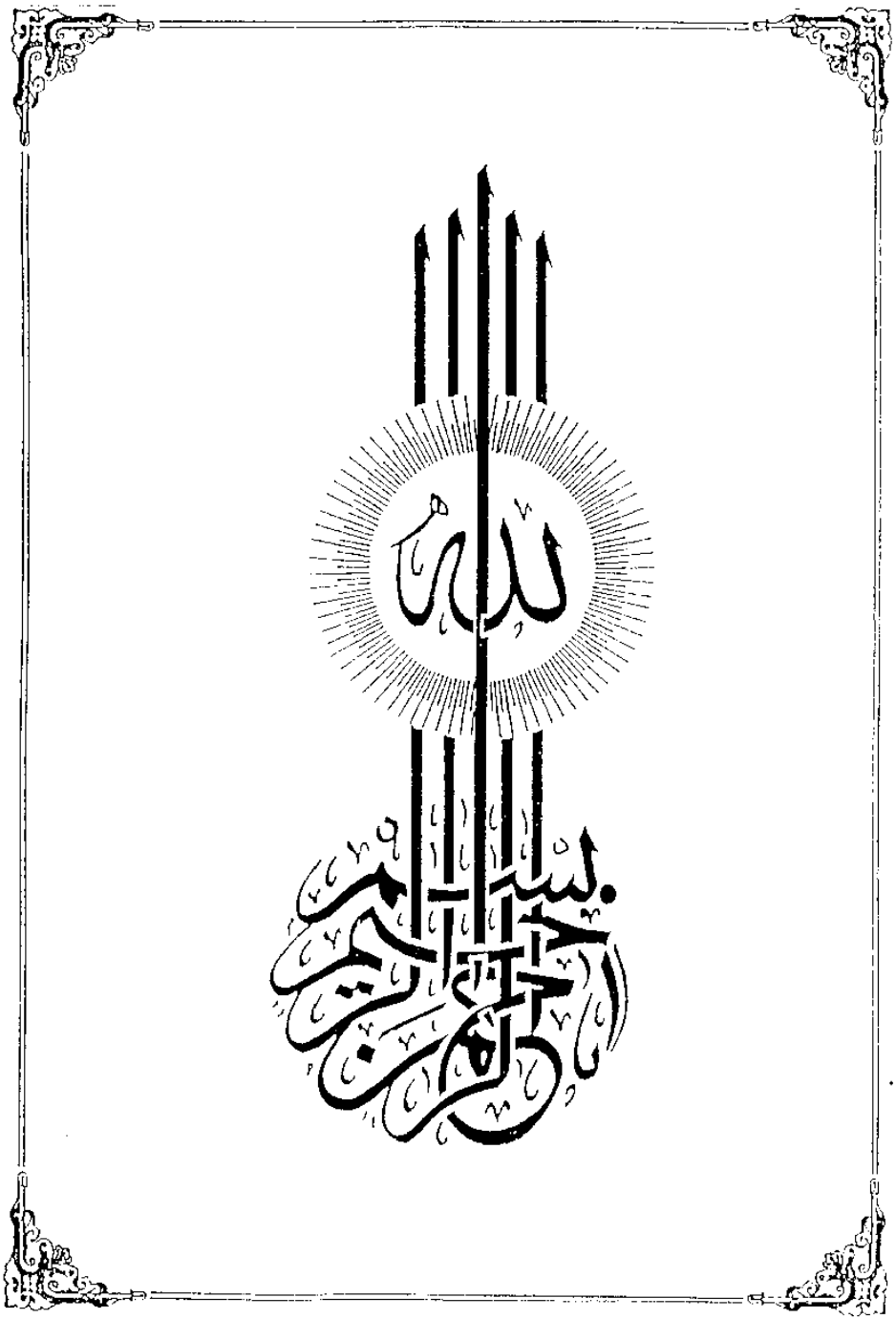
Ali H. Saad



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Ultrasound has become an essential part of ophthalmic diagnosis because of its ability to detect, outline and characterize soft tissues of the eye and orbit with a great percentage of reliability regardless of intervening opacities of the tissues *(Coleman and Dallow, 1985)*.

Since its first military application in 1916, during world war I, the pulse-echo technique of ultrasonography has directed the thoughts of medical scientists to its utilization *(Diamond and Ossoinig, 1980)*.

The phenomenon of resonance of some crystal material, i.e., ceramics as a response to high electrical voltage was discovered in 1880 by the Curies.

An ultrasound system using the "A-mode" as a diagnostic device was designed and fabricated by *Mundt and Hughes in 1956*.

Many investigators have described ultrasonic "A-scan" diagnostic features in early works. Among these *Oksala (1959 and 1964), Bronson (1965 and 1969), Massin and Poujol (1969)* and *Ossoinig (1972)*.

B-scan ultrasonography was first applied to ophthalmology by *Baum and Greenwood in 1958*.

In the late 1960's, Bronson and Turner developed the contact "B-mode" scan used with a lid coupling gel, but this was introduced into practice in 1972 *(Bronson, 1972)*.

At nearly the same year *Singer et al. (1973)* developed a handy, compact and simple apparatus permitting contact and water-immersion B-mode together with A-mode.

The M-mode technique was introduced by *Coleman and Weininger in 1969*.

Since then, many modifications for the instrument were undergone, reducing the size of the machine and providing better resolution. For instance: the vector control scan with B-mode (V C Scan), the 16 degree grey scale, the isometric (D-mode), the computer digital store, the freeze mode, the video recording and others.

These modifications helped in ameliorating a three dimensional interpretation for the results obtained *(Poujol, 1984)*.

FUNDAMENTAL PHYSICS OF ULTRASOUND

Ultrasound waves are those sound waves exhibiting frequencies above 20 Kilohertz [KHz] (20,000 Hz = 20,000 cycles/sec), which is above the audible range of the human ear *(Shammas, 1984)*.

The acoustic wave consists of a sequence of compressions and rarefactions transmitted through a medium. The rate at which the waves are compressed and expanded is called the frequency [f].

Unlike other electromagnetic waves, acoustic waves cannot be transmitted through vacuum.

The Transducer

It is the key element in any ultrasonographic system. It is made essentially of a piezo-electric material which is capable of producing a mechanical displacement when compressed with an electrical voltage. According to this property, when an alternating voltage is applied to the piezo-electric material, the crystalline element will produce alternate compression and expansion waves. The frequency of the generated waves depends on the thickness of the crystalline element.

On the other hand, if the crystal is compressed and expanded by an impinging acoustic wave, a variation in the

molecular charge occurs in the piezo-electric material *(Coleman et al., 1977)*.

It is obvious from this physical property that the transducer crystal material is used for both the generation and detection of sound waves (Fig. 1).

Formerly, the transducer material was made of Quartz, but now, most transdusers are made of more sensitive materials to small voltage difference, including Lithium sulfate and Ceramics such as Lead-Zirconate-Titanate [P-Z-T], these require a relatively minimal voltage difference for excitation (only about 200 Volts) and can detect extremely small ultrasonic signals.

In Figure 2, a diagramatic representation of the basic components of a transducer is shown:

1. The piezo-electric crystal material.
2. An acoustic lens, used to focalize the sound waves.
3. A backing section that acts as a damping space decreasing wave reverbations.

Propagation of sound waves

When a piezo-electric transducer is excited, and the ultrasonic wave is generated, this passes to the coupling material and then to the tissues. Sound wave transmission of air is minimal. The coupling material could be saline solution at body temperature in water-immersion technique, or a gel material in direct contact technique.

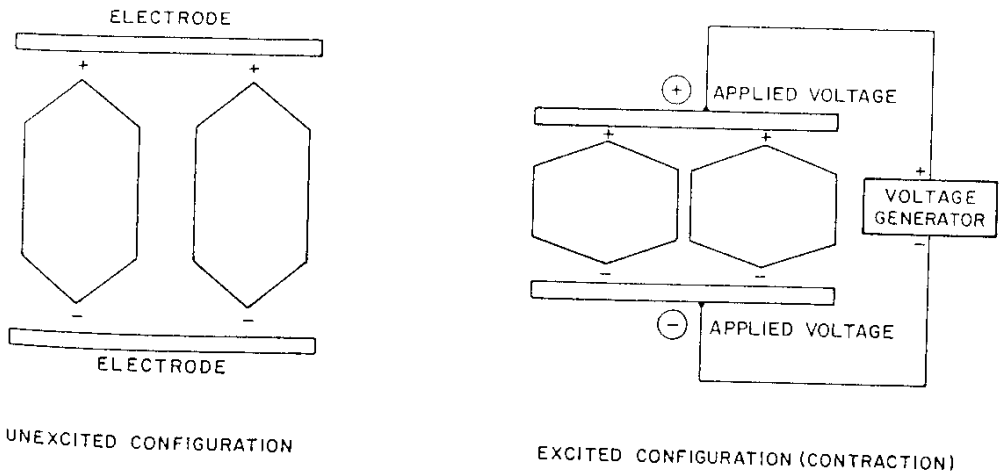

Fig. 1 Schematic representation of molecular configuration in a piezoelectric material illustrating contraction induced by an applied voltage (Coleman et al., 1977).

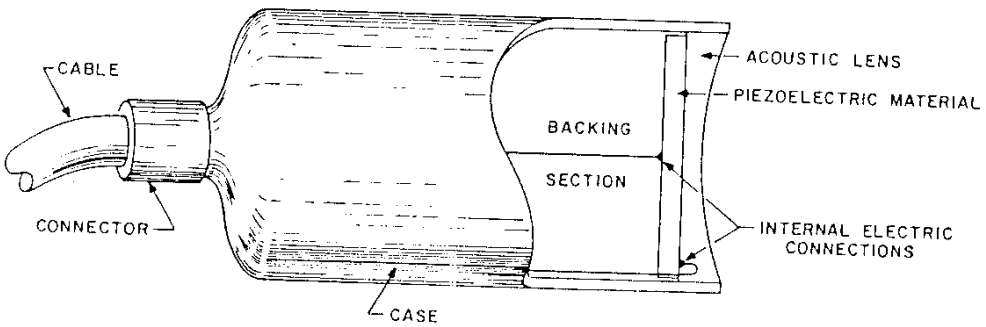

Fig. 2 Cutway view of the transuducer (Coleman et al., 1977).

The waves advance as a series of alternate compressions and rarefactions (pulse-echoes) away from the transducer surface to different tissue depths.

The distance between two successive compressions, or rarefactions is called the wave length [λ]. It represents the length of a complete cycle and is expressed in mm, and since [λ] is small in ophthalmic uses, it can provide beneficial detailed resolution of the tissues.

Velocity [C]

The propagation velocity [C] differs from one tissue to the other, and from one material to the other *(Shammas, 1984)*. It depends upon the density of the material, elasticity and temperature.

Table 1 shows relative propagation velocities for tissues. The mean assumed velocity for soft tissues is about 1540 m/sec and this is the velocity used for our calculations in biometric studies.

This mean velocity is near that of aqueous, vitrous and cornea where the source of error does not exceed 1%.

A bigger source of error is present in cataractous lens 6.1% and in bone 54%. These two structures are better avoided in biometry and in scanning *(Coleman et al., 1977)*.

[C] is directly proportional to [λ] and to the frequency [f]

$$C = \lambda \times f$$

Table [1] showing different propagation velocities of some physiological and pathological constituents of the eye

Substance or tissue	C in m/sec
Water	1524+
Aqueous	1532*
Lens	1641*
Vitrous	1532*
Fat	1476+
Cataractous lens	1629+
Sclera	1630+
Cornea	1550+
Metals (solids)	6000@
Bone	3380@
Muscles	1631+
Optic nerve	1615@

* *Janson and Kock, 1962.*

\+ *Coleman et al., 1977.*

@ *Shammas, 1984.*

Frequency [f]

In diagnostic uses, the higher the frequency the better the resolution, but the less the penetration power as the sound energy is absorbed by the tissues.

The frequency is the number of cycles per second. The unit is the "Hertz" defined as one cycle/second.

Ophthalmic ultrasound uses frequencies ranging from 5 to 20 MHz (1MHz = 1 megahertz = 1 million cycle/second).

For that reason 5 and 8 MHz transducers are better used in orbital diagnosis, 15 MHz transducers show better the anterior segment and 10 MHz transducers are used to show both ocular and orbital structures.

Behavior of sound waves in tissues

Sound waves behave much like light waves and may be altered by external factors following the Snell's law of light. The sound beam has a certain level of energy when it is emitted from the transducer. This level of energy decreases when the ultrasound beam propagates within ocular and orbital tissues. This phenomenon is called *"attenuation"* and results from multiple factors; namely, spreading, absorption, reflection and scattering (*Shammas, 1984*).

Spreading

Ultrasound energy spreads in all directions but the main flow follows the beam axis.

Absorption

A part of the energy is absorbed by the examined tissues and transformed into heat.

Scattering

Diffracted energy by small interfaces.

It should be borne in mind that the sound propagation velocity in solids is larger than fluids and is minimal in gases. So that an acoustic "*converging*" lens of the transducer is a concave and not a convex lens, which is the reverse of a light converging lens as light velocity in solids (lens material, glass) is lower than its velocity in air.

Reflection

A portion of the ultrasound beam is reflected by a large interface, the remainder propagates within deeper tissues. Acoustic waves are reflected at the boundaries between the media, i.e., interfaces and travel partially in the reverse direction to be detected by the transducer as a potential voltage change.

Clinically large interfaces (more than 0.5mm in diameter) are those responsible for detectable reflection or refraction of sound waves.

For the reflection to be maximal, the incident wave front should be as close to the normal to the interface (nearly 90° incidence) and the interface to be regular to prevent sound scatter (Figs. 3 and 4).

The time [t] elapsed from the wave generation by the transducer to its perception is equal to 2 l/C, where [l] is the distance of the interface from the transducer.

$$t = 2\ l/C.$$

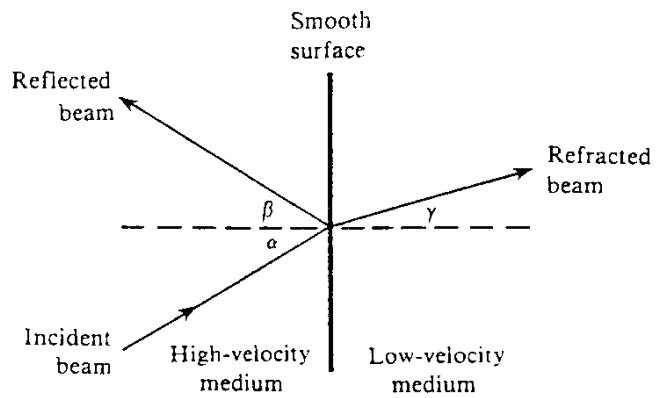

Fig. 3 Changes in a wave front at a smooth surface (Shammas, 1984)

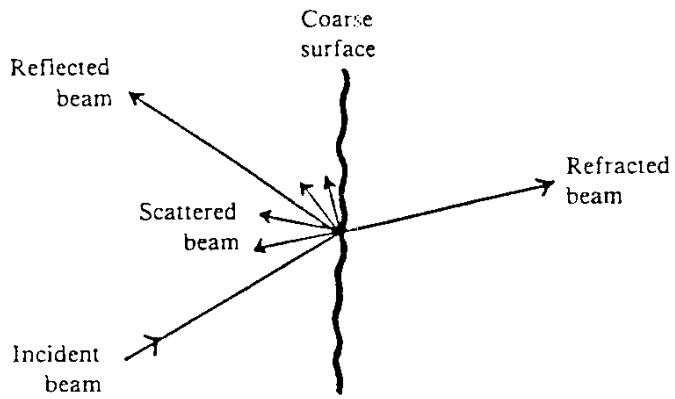

Fig. 4 Changes in a wave front at an irregular surface (Shammas, 1984)