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Sound in medicine

Sound : It is the audible waves of frequency between **20 Hz and 20 kHz**.

Infrasound : refers to the sound of frequency below the

normal hearing range **(<20 Hz)** and subsonic (0 to

20Hz),which cannot be heard.

Ultrasound : It ranges above **20 kHz** ,which is also cannot be heard.

General Properties of Sound

. Sound waves require a medium for their transmission. **. Matter must be present for sound to travel**.

1. *A sound wave* is a mechanical disturbance in a gas ,liquid or solid cause local pressure increase (i.e. compression)and pressure decrease (rarefaction).
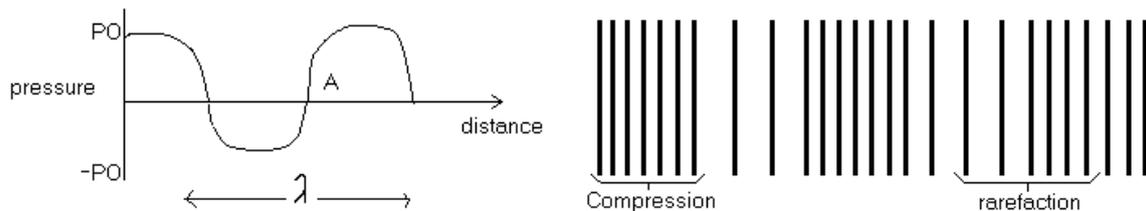

Figure (12.1) Asound wave vibrate at frequency f and produces compression and rarefaction**.**

2. *Sound wave* spread outward as a longitudinal wave i.e. the pressure changes occure in the same direction that the wave travels.

3. *Velocity of the sound* is given by :

$$V = \lambda f$$

f= frequency of vibration of the sound wave.

λ =wave length of the sound wave.

* *The velocity of the sound differs from the medium to medium because* of the freedom of motion of the molecules in the medium, that is their stuffiness elasticity, or compressibility

* *The higher the density, the higher is the velocity of the sound*. Table (12-1). shows the velocity of ultrasound in several substances of medical importance.

Table 12-1. Velocity of ultrasound in several materials of medical interest

Materials	Velocity (m/s)
Air	348
Aluminum	2700
Beryllium	12,890
Blood	1570
Bone	3360
Fat	1500
Liver	1550
Muscle	1580
Oil	1500
Polyethylene	920
Soft tissue	1540
Water	1480

Note that the velocity of ultrasound in bone is twice that in soft tissue and the velocity in soft tissue is five times that in air. *The velocity of ultrasound does not depend on frequency, it is determined by the medium*.i.e..

Speed in gas < speed in liquid < speed in solid

The two types of ultrasound employed in diagnostic are:
1. continuous wave
2. pulsed wave.

Figure (12-2). shows the difference between these types of emission.
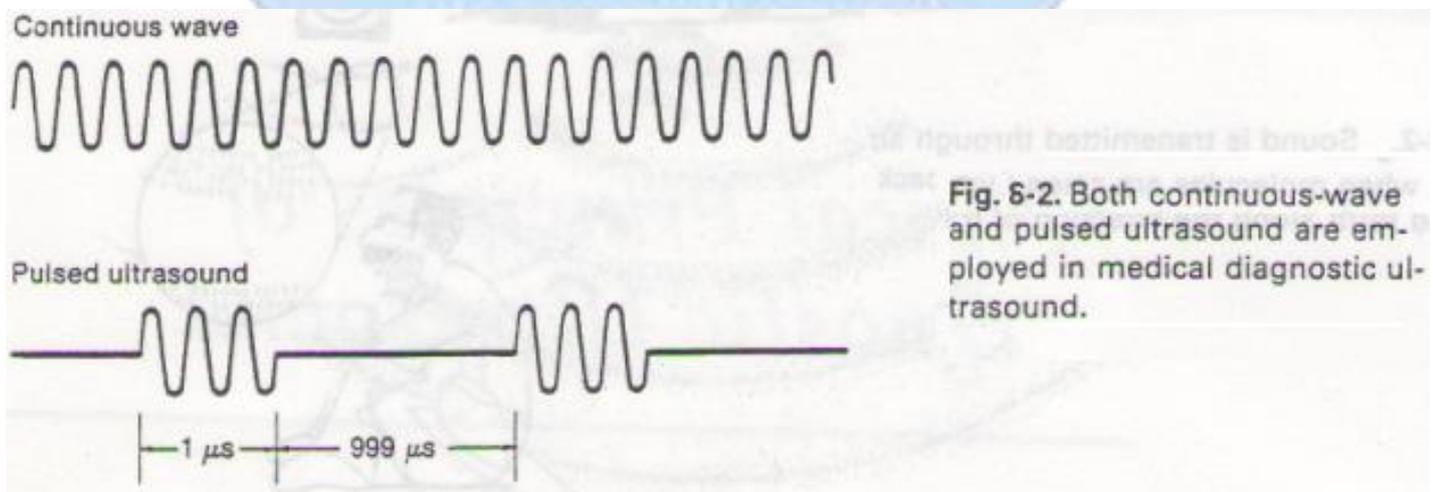

Continuous wave

Pulsed ultrasound

|— 1 μs —|— 999 μs —|

Fig. 6-2. Both continuous-wave and pulsed ultrasound are employed in medical diagnostic ultrasound.

* *During continuous-wave emission*, the ultrasound generating device, *the transducer, vibrates continuously*. This type of ultrasound is principally employed in examinations of the fetal heart and of blood flow by the Doppler method.

* *Pulsed ultrasound* is employed in most imaging studies, including A-mode, B-mode, M- mode, and real time. These imaging studies are based on pulse-echo techniques in which **a** *pulse of ultrasound is emitted and the reflected ultrasound wave or echo, is received by the same transducer after a time delay.*

Acoustic Impedance

Acoustic impedance (**Z**) is used to describe the reflection of sound at an interface.

It is a function of the density of the medium and its compressibility, which is measured by the velocity of sound in the medium. , mathematically, acoustic impedance is described by:

$$Z = \rho V$$

Where ρ : is the density of the medium (Kgm/m^3)

            V : is the velocity of the sound in the medium (m/ s)

∴ Acoustic impedance therefore has unit ($Kgm/m^2.s$)

In general*, the higher the density, the greater is the acoustic impedance. Also , the higher the velocity of sound in the medium, the greater is the acoustic impedance*.

Table (12-2). reports the acoustic impedance for several materials of diagnostic importance since the acoustic impedance is determined by the velocity of sound in the medium, *it is not dependent on the frequency or wavelength of the ultrasound beam.*

Table 12-2. Acoustic impedance for several materials of diagnostic impedance

Materials	Acoustic impedance kg/m^2s (10^{-6})
Air	0.0004
Aluminum	17
Beryllium	1.61
Blood	7.80
Bone	1.58
Fat	1.38
Kidney	1.62
Liver	1.65
Muscle	1.70
Oil	1.43
Polyethylene	1.88
Soft tissue	1.63
Water	1.48

* Acoustic impedance is a most important tissue characteristic. *The largest reflections occur between tissues with great differences in acoustic impedance.*

The other two parameters constituting the wave equation-frequency and wavelength are inversely proportional. As ultrasound frequency increases, the wavelength decreases.

The ability to resolve small objects is directly related to the wavelength of the radiation involved.

* *High –frequency ultrasound (short-wavelength) results in better resolution than the low frequency.*

* High-frequency ultrasound results in shallow penetration. For this reason, high-frequency transducers, up to **15 MHz**, are employed for ultrasonic examination of *small superficial structures such as eyes*.
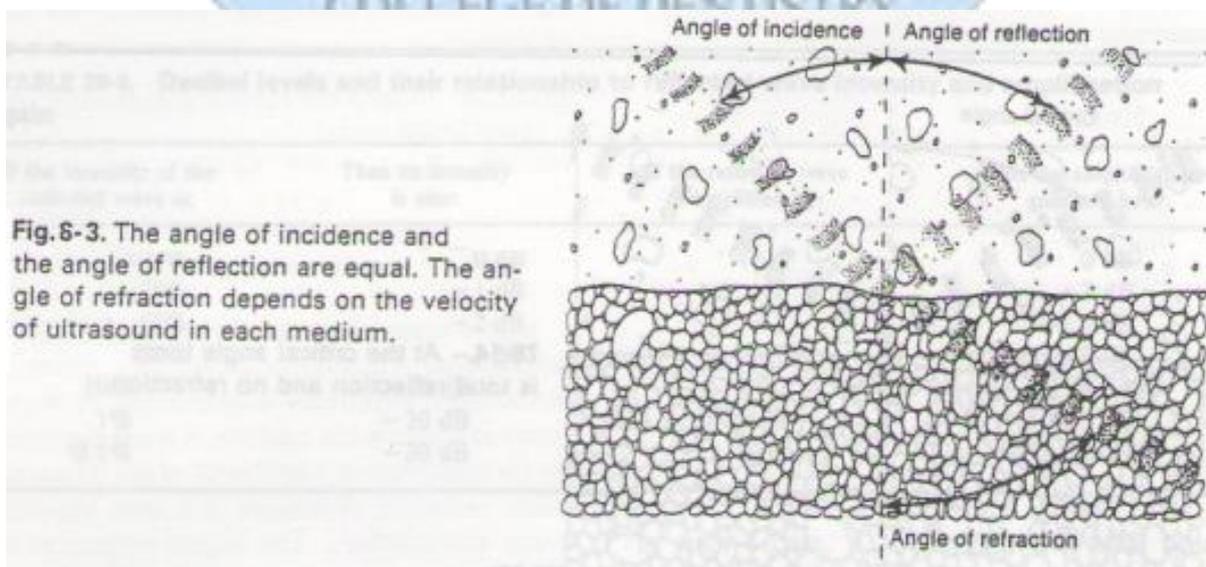
* Lower frequency transducers, around **2.5 MHz** , are used for *abdominal* ultrasound examinations.

Diagnostic ultrasound is usually identified by its frequency of operation. In that regard, some summary statements are possible. As the frequency of ultrasound increases, the following occurs:

1. The ability to resolve small objects improves.
2. The penetrability of the beam decreases.
3. The beam becomes more collimated and directional.

Reflectivity

When an ultrasound wave incident on a tissue interface, some of the sound will be reflected and some will be transmitted figure (12-3). The transmitted beam will leave the interface at an angle different from that of the incident beam. The deviation of the beam is called *refraction*.

Fig. 8-3. The angle of incidence and the angle of reflection are equal. The angle of refraction depends on the velocity of ultrasound in each medium.

** In diagnostic ultrasound the transmitted beam is necessary to image deeper tissues.*

* Principal interest lies in the reflected beam or actually its intensity relative to the incident beam. This quantity of *intensity ratio* is known as the *interface reflectivity* (R) is defined as:

$$R = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2}\right)^2$$

The percentage of ultrasound reflected at such an interface is

$$\text{Percent reflected} = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2}\right)^2 * 100$$

As the angle of incident increases figure (12-4), the reflected beam is directed farther from detecting transducer. As the angle increases still farther, total reflection will occur at the interface, but this reflected beam will note be detected.
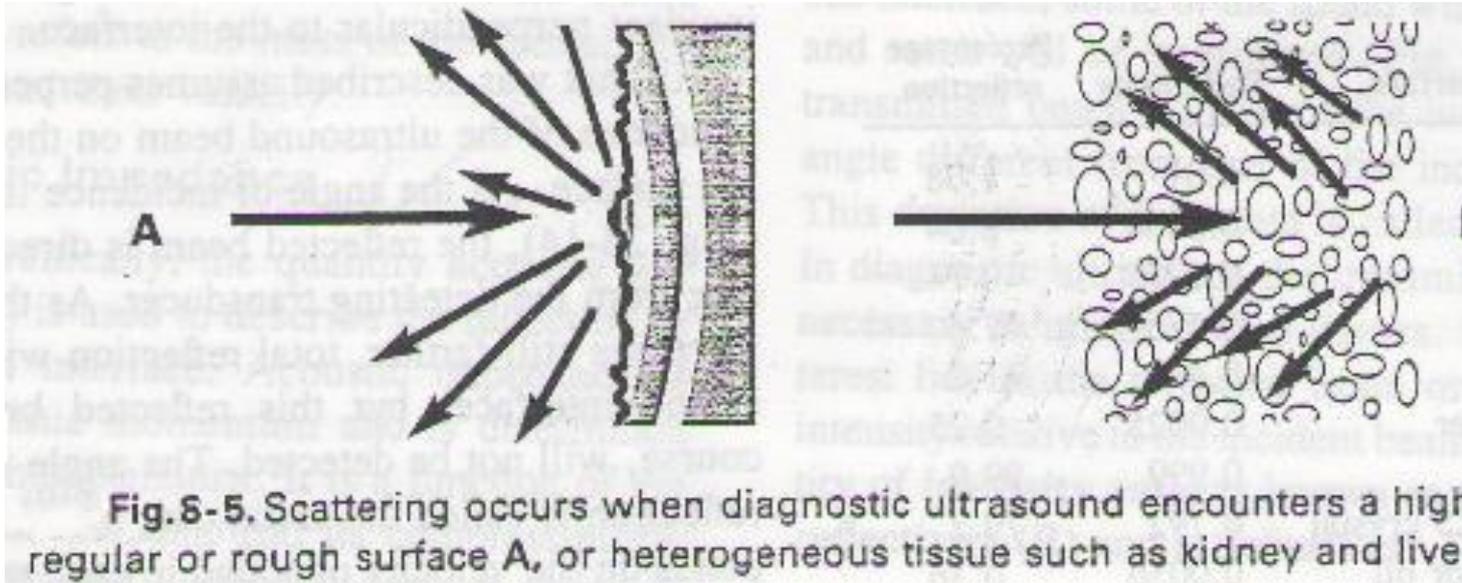

Critical angle

Fig. 8-4. At the critical angle there is total reflection and no refraction.
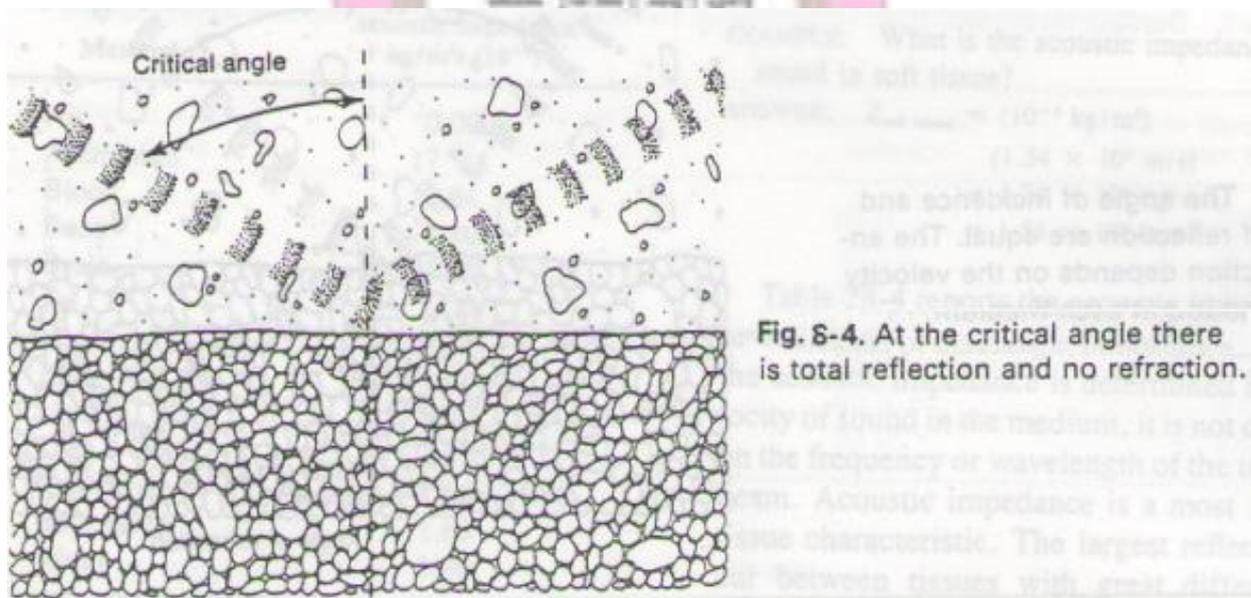
The angle for total reflection is called the critical angle, and *it depends on the velocity of sound in each medium.*

Scattering

Regardless of the angle of incidence, a large fraction of the transmitted ultrasound beam will be reflected if the *roughness* of the tissue interface is large compared to the wavelength of the ultrasound. Such a situation is normal in diagnostic ultrasound and is termed *specular reflection.*

* *if the roughness of the tissue interface is small compared to the wavelength of the transmitted ultrasound, the specular reflection will not occur,* in such a case, the ultrasound beam becomes diffuse and intense because of multiple scattering.

* Highly irregular interfaces and heterogeneous tissues result in scattering, as seen in figure (12-5). Some of the ultrasound is scattered back to the transducer and contributes to image formation. This is call ***backscattered*** ultrasound.

Fig. 8-5. Scattering occurs when diagnostic ultrasound encounters a high regular or rough surface A, or heterogeneous tissue such as kidney and liver

Absorption

Attenuation refers to the reduction in the beam intensity with depth in tissue caused by ***absorption, scattering, and beam divergence***. US attenuation occurs exponentially in much the same manner as that for X-radiation.

The equation describing US attenuation is also similar to:

$$\therefore \qquad I = I_0 e^{-2\alpha X} \ \dots\dots\dots\dots\dots (1)$$

where

  I_0 = is the initial intensity.
  I_x = is the intensity at depth X.
  X = is the depth in tissue.
  α = is the US attenuation coefficient

.

US attenuation coefficient varies with a physical properties of conducting medium and the US frequency.

* The higher frequency, the higher is the attenuation coefficient and therefore the greater attenuation.

For this reason the practical limit to diagnostic US is approximately 15 MHz.

The half – value layer(HVL)=$X_{1/2}$: is the thickness of absorbing tissue will reduce the beam intensity to half it's original value ($I_o/2$) , i.e

at **$X= X_{1/2}$** then **$I= I_o /2$**:

substitute this condition in equation (1) :
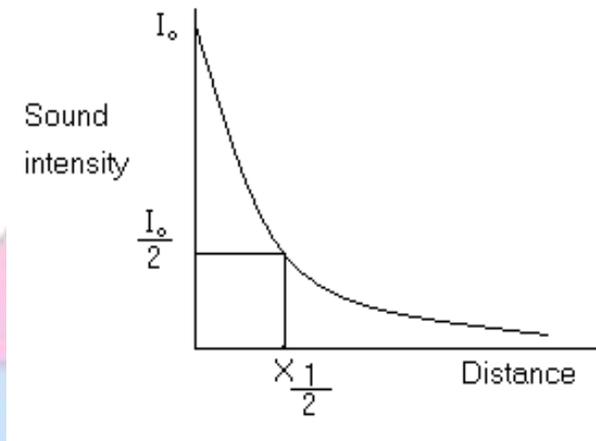
$$I_o /2 = I_0\, e^{-2\alpha X_{1/2}}$$
$$\therefore\ 1/2 = e^{-2\alpha X_{1/2}}$$
$$2^{-1} = e^{-2\alpha X_{1/2}}$$
$$-ln\ 2 = -2\alpha\ X_{1/2}\ ln\ e \qquad since\ ln\ e = ln\ 2.718 \approx 1$$
$$ln\ 2 = 2\alpha\ X_{1/2} \qquad and\ ln\ 2 = 0.693$$

$$X_{1/2} = 0.693 / (2\alpha)$$

<u>*Note*</u> :

    Pulses of US are transmitted into the body by placing the vibrating crystal in close contact with the skin by using a jelly paste or water to :

1- Eliminate the air.

2- Give good coupling at skin.

3- Greatly increase the transmission of US. into the body and of the echoes to the detector (***Transducer***)

Diagnostic Ultrasound Instrumentation and Operation

 Principal reasons for its wide application are its ease of use, the relatively **low cost** of the instrumentation, and the **lack of ionizing radiation.**
* The heart of diagnostic ultrasound is the transducer.

Ultrasound transducer

  A transducer is any device that converts energy from one form to another. An ultrasound transducer converts electric energy into ultrasound energy and ultrasound energy back into electric energy.

* Operation of an ultrasound transducer is based on the piezoelectric effect.
  The piezoelectric effec : is demonstrated graphically in figure (12-6). when a suitable crystalline material is stimulated electrically, the crystal will expand along its short axis. If the polarity of the electric signal is reversed, the crystal will contract.
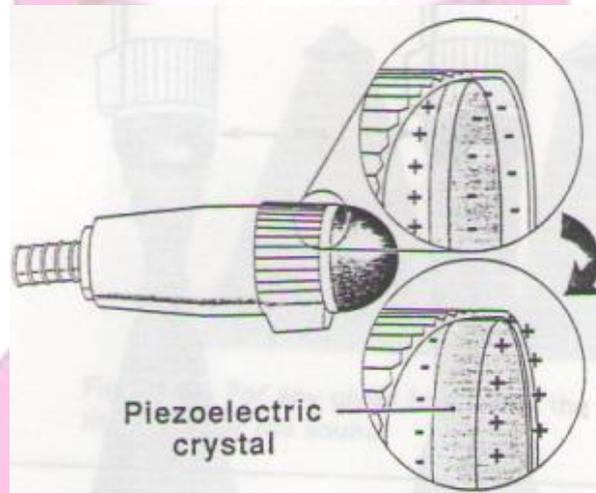

Fig. 6-6. Piezoelectric effect is the distortion of crystal shape by an electrical stimulus and the generation of an electrical signal by mechanical distortion of the crystal.

 If the electric signal oscillates at a high frequency, then the crystal will alternately expand and contract at the same frequency.

* An ultrasound transducer converts an electric signal into mechanical motion and the mechanical motion into ultrasound.

   The reverse is also possible. Ultrasound incident on a suitable crystalline material will transfer the energy of compression and rarefaction into contraction and expansion of the crystal. This in turn will cause an oscillating electric signal. *This process is also the piezoelectric effect.*

Several components comprise the transducer. Figure(12)

The face of the transducer assembly is a protective acoustic window designed to match the active crystal and transmit the ultrasound beam through acoustic coupling to the patient.

A matching layer with acoustic impedance between that of the face and tissue may be attached to improve ultrasound transmission into tissue by reducing surface reflectivity
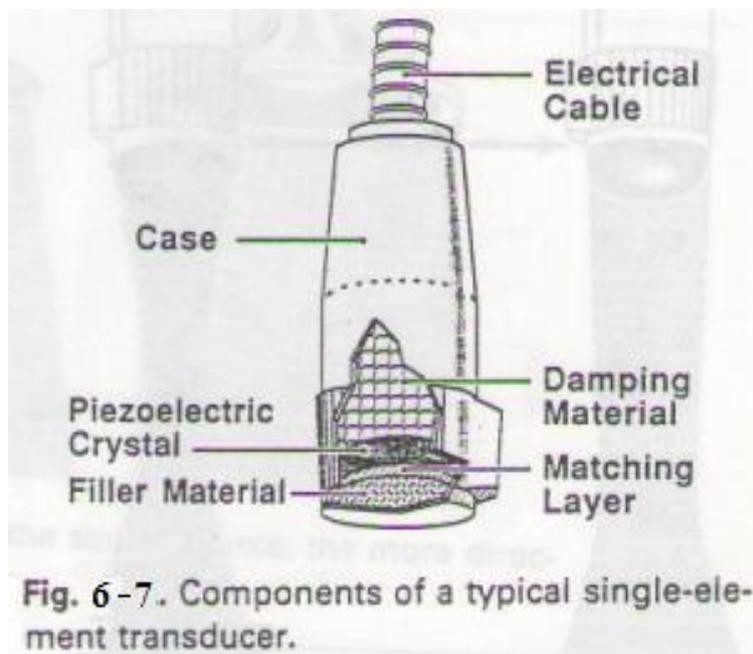

Fig. 6-7. Components of a typical single-element transducer.

* *The active element of the transducer is the piezoelectric crystal.* The material most frequently used is lead zirconate titanate (**PZT**).

* *To optimize the efficiency of US transmission and reception, the thickness of the crystal must be a half or a quarter wavelength.*

For 2.3 MHz transducer, crystal thickness is **0.31mm** or **0.15mm**.
At   10 MHz the crystal thickness is only **70µm** or **35µm.**

* *The piezoelectric crystal* is backed by material designed to damp the movement of the crystal so that, when the electric stimulus is removed, the crystal will cease motion immediately.

* *The piezoelectric crystal and backing material* are surrounded by acoustic insulation to further confine the ultrasound beam. Electric signals are transmitted through a connecter on the back of the trasducer to each of the piezoelectric crystal. The crystal faces are coated with electrically conducting material.

Operational modes

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Two are static imaging modes, A-mode and B-mode;
- *two are dynamic imaging modes,* M-mode and real time;
- *one, Doppler mode, is a ranging mode.*

All find application in diagnostic ultrasound, and each has its own area for special application.

* **A-mode** is particularly useful for measuring midline shifts of the brain.
* **B-mode** is perhaps the one most widely employed, and it is used primarily for abdominal imaging.
* **M-mode** finds its principal application in dynamic imaging of internal structures.
* **Real-time** ultrasound allows for observation of structures in motion.
* **Doppler** ultrasound is used for depth and flow measurements and investigations of moving surfaces. It finds principal application in fetal heart monitoring and peripheral blood flow measurement.

A-Mode Display

A-mode (amplitude mode), In this mode, ultrasound is emitted in pulses by the transducer, which then also receives echoes or reflections from tissue interfaces. This type of ultrasound emission is called *pulse-echo technique.*

A-mode ultrasound employs one or tow transdure.
* *In a two transducer application*, one transducer is used to transmit and the other to receive. For observation in the brain, the transmitting transducer is placed on one side of the head and the receiving transducer on the other.

* *In the one-device mode*, the same transducer is used to transmit and device.

* *The main purpose for employing A-mode* is to measure the depth of interfaces and to detect their separation accurately.

B-Mode Display

B-mode **(brightness mode)** : has little application in diagnostic ultrasound**.**

B-mode transducer is manipulated, this transducer can be moved linearly over the patient to provide a rectangular field of view, it can be angulated to provide a sector field of view, or a combination of both can be employed

M-Modest Display

This type of ultrasound display is called **M-mode (**motion mode**).** It is also sometimes referred to as **TM-mode** (time-motion mode), **PM-mode** (position-motion mode), and **UCH** (ultrasonic cardiograph), since its principal application is to monitor the heart.

Doppler Ultrasound

Another method of ultrasonically monitoring the movement of tissue interface is based on the *Doppler effect*. The wavelength of light varied according to the relative motion of the source of light and the observer.

* If the source or observer or both are moving toward each other, the light received will have a shorter wavelength (higher frequency) than the emitted. On the other hand, if the source and observer are moving apart, the received light will have a lowerfrequency than that emitted.

A continuous ultrasound beam is emitted in Doppler applications. When the reflected beam is received by the transducer, the change in frequency caused by the Doppler effect is electronically determined.

The mathematics associated with Doppler ultrasound to measure this frequency shift are rather simple.
The frequency shift, F_D is called the Doppler shift frequency.
If F_T is the transmitted frequency and F_R the frequency of sound reflected from the moving tissue interface, then

$$F_D = F_T - F_R$$

If however, one knows the velocity of sound in the medium (V) and the velocity in the interface (u), then the Doppler shift frequency becomes:

$$F_D = F_T \frac{2u}{V}$$

The transducer used in Doppler US incorporates two crystals, one to transmit and one to receive. Because of this simplicity, the cost of a Doppler ultrasound unit can be much less than that of a compound B-scanner.

Real-Time Imaging

Real-time ultrasound is dynamic imaging. It is to compound B-mode what fluoroscopy is to radiography. Real-time ultrasound is finding increasing application in many areas of medical imaging.

It has several distinct advantages over B-mode imaging:
1. The cost of equipment can be considerably less.
2. The image obtained is not nearly so dependent on operator skill.
3. The time required for real-time examination is generally less because of the ease with which the equipment can be handled.
4. Several commercial versions are available, including mobile system.

The real-time transducer assembly is larger than a B-mode transducer.

The transducer probe* is then moved over the surface of the patient in any direction and angle until the anatomic region of interest is found

The dynamic (moving) image may then be stored on videotape for subsequent viewing, or stop-action frame photographs may be obtained.

Real – time ultrasound does have disadvantages.

The real-time image results from the ultrasound beam interacting with the tissue interface from only one direction, whereas with B-mode, one can move the transducer while storing the image from many directions for ultimate composition.

Consequently, *the resolution is better with B-mode than with real time.*

BIOLOGIC EFECT

Mechanisms of Action

An expression used by radiologists to describe the manner in which radiation produces a biologic effect is **the mechanism of action.**
* *For ionizing radiation*, the mechanism of action is ionization and excitation.
* *For ultrasound*, the mechanism of action is temperature elevation, cavitations and various viscous stresses.

Thermal effect

Ultrasound irradiation can elevate the temperature of tissue through molecular agitation and the relaxation processes. Extremely intense levels are required to produce a measurable temperature elevation in tissue.

* The hazard from temperature elevation is, of course, not specific to ultrasound. At the local tissue level, temperature elevation can result in structural changes in macromolecules and membranes and changes in the rates of biochemical reactions.

Effects on Living Tissue

if the ultrasound intensity is sufficiently high, many of the effects described earlier as resulting from ionizing radiation exposure can be produced. *Chemical bonds can be disrupted,* macromolecules are degraded, chromosomes aberration can be produced, and cells can be killed.