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RAD31B
Types And Features Of Waves

Waves come in two kinds, longitudinal and transverse. **Transverse waves** are like those on water, with the surface going up and down, and **longitudinal waves** are like those of sound, consisting of alternating compressions and rarefactions in a medium. The high point of a **transverse wave** is a called the crest, and the low point is called the trough. For longitudinal waves, the **compressions** and **rarefactions** are analogous to the crests and troughs of transverse waves. The distance between successive crests or troughs is called the **wavelength**. The height of a wave is the **amplitude**. How many crests or troughs pass a specific point during a unit of time is called the **frequency**. The **velocity** of a wave can be expressed as the wavelength multiplied by the frequency.

Waves can travel immense distances even though the oscillation at one point is very small. For example, a thunderclap can be heard kilometres away, yet the sound carried **manifests** itself at any point only as minute compressions and rarefactions of the air.
Waves

Mechanical Waves
Requires a medium

- Longitudinal Waves
  - Sound waves
- Transverse Waves
  - Water waves

Electromagnetic Waves
Do not require a medium

- Light
- Microwaves
- Infrared
- X-rays
- Ultraviolet rays
- Radio waves

Figure: Examples of transverse and longitudinal waves.

Transverse Wave

Longitudinal Wave
waves can be in the group and such groups are called **wave packets**, so the velocity with a wave packet travels is called **group velocity**. The velocity with which the phase of a wave travels is called **phase velocity**. The relation between group velocity and phase velocity are proportionate

The group velocity \( v_g \) is defined by the equation:

\[
v_g = \frac{\partial \omega}{\partial k}
\]

where \( \omega \) is the wave's **angular frequency** (usually expressed in radians per second), and \( k \) is the **angular wavenumber** (usually expressed in radians per meter). The **phase velocity** is: \( v_p = \frac{\omega}{k} \).

The function \( \omega(k) \), which gives \( \omega \) as a function of \( k \), is known as the **dispersion relation**.

- If \( \omega \) is **directly proportional** to \( k \), then the group velocity is exactly equal to the phase velocity. A wave of any shape will travel undistorted at this velocity.
- If \( \omega \) is a linear function of \( k \), but not directly proportional \( (\omega = ak + b) \), then the group velocity and phase velocity are different. The envelope of a **wave packet** will travel at the group velocity, while the individual peaks and troughs within the envelope will move at the phase velocity.
- If \( \omega \) is not a linear function of \( k \), the envelope of a wave packet will become distorted as it travels. Since a wave packet contains a range of different frequencies (and hence different values of \( k \)), the group velocity \( \frac{\partial \omega}{\partial k} \) will be different for different values of \( k \). Therefore, the envelope does not move at a single velocity, but its wavenumber components \( (k) \) move at different velocities, distorting the envelope.
Ultrasound Principles

Sound waves are produced by vibrating sources, which cause particles in the medium to oscillate back and forth, setting up the propagating pressure wave. A wave is a traveling variation in something, such as pressure in the case of sound. As sound propagates, it is attenuated (weakened), scattered (spread out), and reflected (bounced back), producing echoes from anatomic structures. In medical ultrasonography, the transducer serves as the source and receiver of sound waves. Transducers are designed such that the sound waves they generate travel in a narrow beam with a well-defined direction. The reception of reflected and scattered echo signals by the transducer not only produces ultrasound images but also allows the detection and measurement of motion using the Doppler effect. This section discusses factors that are important in the transmission and reflection of ultrasound in tissue.

Speed of sound

Most ultrasound applications involve transmitting short bursts, or pulses, of sound (typically two or three cycles long) into the body and receiving echoes from tissue interfaces. The time between transmitting a pulse and receiving an echo is used to determine the depth of the interface. The speed of sound in tissue must be known in order to calculate this depth.

The velocity of sound waves mostly depends on the properties of the transmitting medium and not significantly on the frequency or the wave amplitude (strength). As a general rule, gases, including air, exhibit the lowest propagation velocities, liquids have an intermediate range of velocities, and solids have
the highest sound transmission velocities. For soft tissues, the average speed of sound is 1.54 mm/µs (1540 m/s). Variations exist in the speed of sound from one tissue to another, but, as in given table indicates, the speed of sound in various soft tissues deviates only slightly from the assumed average. On average, the speed of sound transmission in fat is lower than that in muscle.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Speed of Sound (mm/µs)</th>
<th>Percentage Change From Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fat</td>
<td>1.45</td>
<td>−5.8</td>
</tr>
<tr>
<td>Vitreous humor</td>
<td>1.52</td>
<td>−1.3</td>
</tr>
<tr>
<td>Liver</td>
<td>1.55</td>
<td>+0.6</td>
</tr>
<tr>
<td>Blood</td>
<td>1.57</td>
<td>+1.9</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.58</td>
<td>+2.6</td>
</tr>
<tr>
<td>Lens of eye</td>
<td>1.62</td>
<td>+5.2</td>
</tr>
<tr>
<td>Soft tissue average</td>
<td>1.54</td>
<td></td>
</tr>
</tbody>
</table>
Radiation Physics

Department of radiological technique

Third stage
A Longitudinal Wave — A wave for which the direction of displacement for the medium’s particles is parallel to the direction of wave propagation.

Transverse Wave — A wave for which the direction of displacement for the medium’s particles is perpendicular to the direction of wave propagation.

This basic division is frequently used in ultrasound, but it is important to note that there are also wave types for which the direction of motion for the medium’s particles is not fixed relative to the wave propagation direction. For example, in surface waves (as in sea waves) the angle between the two directions changes continuously. A second and important division is based on the wave front geometry. Using this
approach we can again divide the waves into two basic types.

A. A Planar Wave — A wave for which the wave front is located on a plane that propagates in space.

B. A Circular Wave — A wave that propagates symmetrically around a reference point (as a sphere or a ring), or around a reference line (as a cylinder).

Frequency and wavelength

The number of oscillations (cycles) per second of the vibrating elements in the transducer is the frequency of the sound wave. Frequency is expressed in cycles per second, or hertz (Hz). Audible sounds are in the range of 20 Hz to 20 kHz. Ultrasound refers to sound whose frequency is above the audible range (the prefix ultra means “beyond”). Diagnostic ultrasound applications use frequencies in the 2 to 15 MHz (mega hertz, i.e., 2 million to 15 million Hz) frequency range. Because higher frequencies are associated with improved spatial detail (i.e., better detail resolution), sonographers use the highest
frequency that still allows adequate depth to visualize tissue in a given scanning situation. Some applications that require very little penetration use frequencies as high as 50 MHz.

Fig. 2.1 shows a sound wave frozen in time. It illustrates accompanying compressions and rarefactions (expansions) in the medium that result from the pressure and particle oscillations. The wavelength $\lambda$ is the spatial length of a cycle where $c$ is the speed of sound and $f$ is the frequency. Table 2.2 presents values of wavelengths in soft tissue for several frequencies. For soft tissues, the wavelength $\lambda_t = \frac{1.54 \, \text{mm}}{f}$, where $f$ is the frequency expressed in megahertz. For example, if the frequency is 5 MHz, the wavelength in soft tissue is approximately 0.3 mm. Higher frequencies have shorter wavelengths leading to shorter pulses and improved detail resolution.
Sound waves produced by an ultrasound transducer. Vibrations of the transducer are coupled into the medium, producing local fluctuations in pressure. The fluctuations propagate through the medium in waves. The pressure amplitude is the maximum pressure variation, positive or negative. The diagram schematically illustrates compressions and rarefactions at an instant of time. The symbol \( \lambda \) is the acoustic wavelength \( c = \lambda f \).

**TABLE 2.2**

<table>
<thead>
<tr>
<th>Frequency (MHz)</th>
<th>Wavelength (mm) Assuming 1.54 mm/µs</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>0.77</td>
</tr>
<tr>
<td>5</td>
<td>0.31</td>
</tr>
<tr>
<td>10</td>
<td>0.15</td>
</tr>
<tr>
<td>15</td>
<td>0.10</td>
</tr>
<tr>
<td>20</td>
<td>0.08</td>
</tr>
</tbody>
</table>

Wavelength has relevance when describing dimensions of anatomic structures. The size of an object is more easily understood if given relative to the ultrasonic wavelength for the frequency of the sound beam. Similarly, the width of the ultrasound beam from a transducer depends in part on the wavelength. Higher frequency beams have shorter wavelengths and can be focused more tightly than lower frequency beams.
Amplitude, intensity, and power

A sound wave is a propagating pressure variation. The pressure profile that occurs for the wave in Fig. 2.1 appears in the lower part of the figure. The pressure amplitude is the maximal increase (or decrease) in the pressure, relevant to normal pressure, caused by the sound wave. The unit for pressure is the pascal (Pa). Ultrasound instruments can produce peak pressure amplitudes of millions of pascals in water when power controls on the instrument are set at maximum. Ultrasound beams to enhance echo signals. Diagnostic levels, however, are not. As a benchmark for comparison, atmospheric pressure is approximately 0.1 MPa, so it is clear that ultrasound beams from medical devices significantly exceed this value. The high-pressure amplitudes of ultrasound pulses can burst contrast agent bubbles (see later and Chapter 35) that are sometimes injected into the believed to create biologic effects in tissues if such gas bodies are not present.

The intensity (I) of a sound wave at a point in the medium is estimated by squaring the pressure amplitude (P) and using \( I = \frac{P^2}{2 \rho c} \), where \( \rho \) is the density of the medium and \( c \) is the speed of sound in it. Units for ultrasound intensity are watts per meter squared (W/m\(^2\)) or multiples thereof, such as mW/cm\(^2\). In water, a 2 MPa amplitude during the pulse corresponds to a pulse average intensity of 133 W/cm\(^2\)! This is a high intensity, but, fortunately, it is not sustained by a diagnostic ultrasound device because the duty factor (i.e., the fraction of time the transducer actually emits ultrasound) is a few percent at most. Therefore the time-averaged acoustic intensity from an ultrasound instrument, found by averaging over a time that includes transmit pulses as well as the time between pulses, is much lower than
the intensity during the pulse. Typical time-averaged intensities at the location in the ultrasound beam where the maximal values are found are in the order of 10 to 20 mW/cm$^2$ for anatomic imaging. Doppler and color Doppler imaging modes have higher duty factors. Moreover, these modes tend to concentrate the acoustic energy into smaller areas. Output data for sonographic instruments are published and must be compliant with predefined limits. The acoustic power produced by an instrument is the rate at which energy is emitted by the transducer. Average acoustic power levels in diagnostic ultrasonography are low because of the small duty factors used in most equipment. Typical power levels of 10 to 20 mW for B-mode imaging can triple or quadruple for Doppler modes of operation.

**PIEZOELECTRIC EFFECT**

The piezoelectric (pressure-electric) effect is a phenomenon in which a material, upon the application of an electrical field, changes its physical dimensions and vice versa (Cady, 1964; Kino, 1987). The piezoelectric effect was discovered by French physicists Pierre and Jacques Curie in 1880. The direct and reverse piezoelectric effects are illustrated in Figure 3.1(a) and (b), respectively. The direct effect refers to the phenomenon in which the application of a stress causes a net charge to appear across the electrodes and the inverse effect concerns the production of a strain upon the application of a potential difference across the electrodes. Certain naturally occurring crystals such as quartz and tourmaline are
piezoelectric. The physical reason that the piezoelectric phenomenon occurs can be idealistically explained by considering that a piezoelectric material consists of innumerable electric dipoles. When undisturbed, these dipoles are randomly distributed, resulting in a neutral state or no net charge. An electrical potential difference applied across a slab of piezoelectric material realigns the dipoles in the material in a preferential direction and results in a deformation or a change in the thickness of the slab. Conversely, a stress that causes a deformation of the material and reorientation of the dipoles induces a net charge across the electrodes.

1 (a) Direct piezoelectric effect in which a stress induces a charge separation. (b) Reverse piezoelectric effect in which a potential difference across the electrodes induces a strain
(a) Direct piezoelectric effect in which a stress induces a charge separation. (b) Reverse piezoelectric effect in which a potential difference across the electrodes induces a strain
Direct Piezoelectric Effect

Reverse Piezoelectric Effect
Naturally occurring piezoelectric crystals are seldom used today as transducer materials in diagnostic ultrasonic imaging because of their weak piezoelectric properties. The most popular material is a polycrystalline ferroelectric ceramic material, lead zirconate titanate, Pb(Zr, Ti)O3 or PZT, which possesses very strong piezoelectric properties following polarization. Polarization of a ferroelectric material is carried out by heating it to a temperature just above the Curie temperature of the material and then allowing it to cool slowly in the presence of a strong electric field; this is typically in the order of 20 kV/cm applied in the direction in which the piezoelectric effect is required. The electrical field is usually applied to the material by means of two electrodes. This process aligns the dipoles along the direction of polarization.

A great variety of ferroelectric materials exists. Barium titanate (BaTiO3) was among the first developed. Lead metaniobate (PbNb2O6) and lithium niobate (LiNbO3) have also been used. Certain piezoelectric properties of PZT can be enhanced by doping. As a result, many types of PZT are commercially available. In order to define and better understand the physical meaning of the piezoelectric properties of a material, the constitutive equations that govern the piezoelectric effect must be examined.
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Third stage
PIEZOELECTRIC EFFECT

The piezoelectric effect is exhibited by certain crystals that, in response to applied pressure, develop a voltage across opposite surfaces. This effect is used to produce an electrical signal in response to incident ultrasound waves. The magnitude of the electrical signal varies directly with the wave pressure of the incident ultrasound. Similarly, application of a voltage across the crystal causes deformation of the crystal—either compression or extension depending upon the polarity of the voltage. This deforming effect, termed the converse piezoelectric effect, is used to produce an ultrasound beam from a transducer.

Many crystals exhibit the piezoelectric effect at low temperatures, but are unsuitable as ultrasound transducers because their piezoelectric properties do not exist at room temperature. The temperature above which a crystal’s piezoelectric properties disappear is known as the Curie point of the crystal. A common definition of the efficiency of a transducer is the fraction of applied energy that is converted to the desired energy mode. For an ultrasound transducer, this definition of efficiency is described as the electromechanical coupling coefficient $k_c$. If mechanical energy (i.e., pressure) is applied, we obtain
\[ k^2 \text{c} = \text{Mechanical energy converted to electrical energy}/(\text{Applied mechanical energy}) \]

If electrical energy is applied, we obtain

\[ k^2 \text{c} = \text{Electrical energy converted to mechanical energy}/(\text{Applied electrical energy}) \]

Values of \( k \text{c} \) for selected piezoelectric crystals are listed in following Table. Essentially all diagnostic ultrasound units use piezoelectric crystals for the generation and detection of ultrasound. A number of piezoelectric crystals occur in nature (e.g., quartz, Rochelle salts, lithium sulfate, tourmaline, and ammonium dihydrogen phosphate [ADP]). However, crystals used clinically are almost invariable man-made.
Properties of Selected Piezoelectric Crystals

<table>
<thead>
<tr>
<th>Material</th>
<th>Electromechanical Coupling Coefficient $K_c$</th>
<th>Curie Point ($^\circ$C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quartz</td>
<td>0.11</td>
<td>550</td>
</tr>
<tr>
<td>Rochelle salt</td>
<td>0.78</td>
<td>45</td>
</tr>
<tr>
<td>Barium titanate</td>
<td>0.30</td>
<td>120</td>
</tr>
<tr>
<td>Lead zirconate titanate (PZT-4)</td>
<td>0.70</td>
<td>328</td>
</tr>
<tr>
<td>Lead zirconate titanate (PZT-5)</td>
<td>0.70</td>
<td>365</td>
</tr>
</tbody>
</table>

The most common man-made crystals are barium titanate, lead metaniobate, and lead zirconate titanate (PZT).

ULTRASOUND INTENSITY

Ultrasound frequencies of 1 MHz and greater correspond to ultrasound wavelengths less than 1 mm in human soft tissue. As an ultrasound wave passes through a medium, it transports energy through the medium. The rate of energy transport is known as “power.” Medical ultrasound is produced in beams that are usually focused into a small area, and the beam is described in terms of the power per unit area, defined as the beam’s “intensity.” The relationships among the quantities and units pertaining to intensity are summarized (Energy (E) ability to do work, Power(P) rate of energy transported. And Intensity(I) is Power per unit...
aria (a) so \( I = P/a \) or \( I = E/ta \), \( t \) is time). Intensity is usually described relative to some reference intensity. For example, the intensity of ultrasound waves sent into the body may be compared with that of the ultrasound reflected back to the surface by structures in the body. For many clinical situations the reflected waves at the surface may be as much as a hundredth or so of the intensity of the transmitted waves. Waves reflected from structures at depths of 10 cm or more below the surface may be lowered in intensity by a much larger factor. A logarithmic scale is most appropriate for recording data over a range of many orders of magnitude. In acoustics, the decibel scale is used, with the decibel defined as

\[
\text{dB} = 10 \log \frac{I}{I_0}
\]  

where \( I_0 \) is the reference intensity. Following table shows examples of decibel values for certain intensity ratios. Several rules can be extracted from this table:

Positive decibel values result when a wave has a higher intensity than the reference wave; negative values denote a wave with lower intensity.

Increasing a wave’s intensity by a factor of 10 adds 10 dB to the intensity, and reducing the intensity by a factor of 10 subtracts 10 dB.

Doubling the intensity adds 3 dB, and halving subtracts 3 dB.

No universal standard reference intensity exists for ultrasound. Thus the statement “ultrasound at 50 dB was used” is nonsensical. However, a statement such as “the returning echo was 50 dB below the transmitted signal” is informative. The transmitted signal then becomes the reference
intensity for this particular application. For

<table>
<thead>
<tr>
<th>Ratio of Ultrasound Wave Parameters</th>
<th>Intensity Ratio $(I/I_0)$ (dB)</th>
<th>Amplitude Ratio $(A/A_0)$ (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1000</td>
<td>30</td>
<td>60</td>
</tr>
<tr>
<td>100</td>
<td>20</td>
<td>40</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
<td>20</td>
</tr>
<tr>
<td>2</td>
<td>3</td>
<td>6</td>
</tr>
<tr>
<td>1</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>1/2</td>
<td>-3</td>
<td>-6</td>
</tr>
<tr>
<td>1/10</td>
<td>-10</td>
<td>-20</td>
</tr>
<tr>
<td>1/100</td>
<td>-20</td>
<td>-40</td>
</tr>
<tr>
<td>1/1000</td>
<td>-30</td>
<td>-60</td>
</tr>
</tbody>
</table>
audible sound, a statement such as “a jet engine produces sound at 100 dB” is appropriate because there is a generally accepted reference intensity of 10⁻¹⁶ W/cm² for audible sound. A 1-kHz tone (musical note C one octave above middle C) at this intensity is barely audible to most listeners. A 1-kHz note at 120 dB (10⁻⁴ W/cm²) is painfully loud. Because intensity is power per unit area and power is energy per unit time Eq. (1) may be used to compare the power or the energy contained within two ultrasound waves. Thus we could also write

\[
\text{dB} = 10 \log \frac{\text{Power}}{P_o} = 10 \log \frac{E}{E_o}
\]

Ultrasound wave intensity is related to maximum pressure (Pm) in the medium by the following expression
\[ I = \frac{P_m^2}{2\rho c} \quad \ldots\ldots\ldots 2 \]

where \( \rho \) is the density of the medium in grams per cubic centimeter and \( c \) is the speed of sound in the medium. Substituting Eq. -2) for \( I \) and \( I_0 \) in Eq. (-1) yield

\[ dB = 10 \log \left( \frac{P_m}{P_{mo}} \right)^2 \]
\[ dB = 20\log\left(\frac{p_m}{p_{mo}}\right) \quad \ldots\ldots\ldots 3 \]

When comparing the pressure of two waves, Eq. (-3) may be used directly. That is, the pressure does not have to be converted to intensity to determine the decibel value. An ultrasound transducer converts pressure amplitudes received from the patient (i.e., the reflected ultrasound wave) into voltages. The amplitude of voltages recorded for ultrasound waves is directly proportional to the variations in pressure in the reflected wave.
The decibel value for the ratio of two waves may be calculated from Eq. (1) or from Eq. (3), depending upon the information that is available concerning the waves. The “half-power value” (ratio of 0.5 in power between two waves) is –3 dB, whereas the “half-amplitude value” (ratio of 0.5 in amplitude) is –6 dB. This difference reflects the greater sensitivity of the decibel scale to amplitude compared with intensity values.

ULTRASOUND VELOCITY

The velocity of an ultrasound wave through a medium varies with the physical properties of the medium. In low-density media such as air and other gases, molecules may move over relatively large distances before they influence neighboring molecules. In these media, the
velocity of an ultrasound wave is relatively low. In solids, molecules are constrained in their motion, and the velocity of ultrasound is relatively high. Liquids exhibit ultrasound velocities intermediate between those in gases and solids. With the notable exceptions of lung and bone, biologic tissues yield velocities roughly similar to the velocity of ultrasound in liquids. In different media, changes in velocity are reflected in changes in wavelength of the ultrasound waves, with the frequency remaining relatively constant. In ultrasound imaging, variations in the velocity of ultrasound in different media introduce artifacts into the image, with the major artifacts attributable to bone, fat, and, in ophthalmologic applications, the lens of the eye. The velocities of ultrasound in various media are listed in Tabl

<table>
<thead>
<tr>
<th>Nonbiologic Material</th>
<th>Velocity (m/sec)</th>
<th>Biologic Material</th>
<th>Velocity (m/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acetone</td>
<td>1174</td>
<td>Fat</td>
<td>1475</td>
</tr>
<tr>
<td>Air</td>
<td>331</td>
<td>Brain</td>
<td>1560</td>
</tr>
<tr>
<td>Aluminum (rolled)</td>
<td>6420</td>
<td>Liver</td>
<td>1570</td>
</tr>
<tr>
<td>Ethanol</td>
<td>1207</td>
<td>Spleen</td>
<td>1570</td>
</tr>
<tr>
<td>Acrylic plastic</td>
<td>2680</td>
<td>Muscle</td>
<td>1580</td>
</tr>
</tbody>
</table>
Radiation Physics

Department of radiological

Third stage

Rad35
What is an ultrasound transducer?

An ultrasound transducer is the handheld device that the technician or doctor moves on or over the body of the patient. A cord connects it to a computer. The device sends sound waves and receives the echoes as they bounce off the body tissue and organs of the patient. These echoes are used by the computer to create an image.

Transducers (probes) come in different sizes and shapes for use in different parts of the body. Some are placed on or over the body part. Others are designed to be inserted into an opening like the vagina or rectum so they can get closer to the organ for a more detailed view.

Ultrasound or sonography is based on the same basic principle used by bats. An ultrasound machine measures the echoes bouncing back to the transducer from the body of the patient to form an image. Bats ‘hear’ echoes and measure them to determine how far away the object is that caused the echo. They use what’s called ‘echolocation’ to fly around at night.
without bumping into anything.
In the transducer probe are piezoelectric crystals that change shape when an electrical current is applied to them. The vibrations or shape changes create sound waves that move outward. When they are directed at the human body, they pass right through the skin and into the internal anatomy.
As the waves encounter tissues with different characteristics and densities, they produce echoes that reflect back to the

---

**Ultrasound Transducer**

*Converts electricity to sound and vice versa*

- Plastic housing
- Co-axial cable
- Matching layer
- Backing block
- Acoustic insulator
- Piezoelectric crystal
crystals. This happens more than a thousand times a second. The returning echoes are converted to electrical signals, and the computer uses them as points of brightness on the image, corresponding to the anatomic position and strength of the reflecting echoes.

A transducer contains a large array of crystals which allow it to make a series of image lines that together form a complete image frame called a sonogram. All the crystals are repeatedly activated many times in such a way that a complete image frame is formed around 20 times per second. This means that ‘real-time- motion is displayed in the ultrasound image.

As images are captured in real-time, they can show how the blood is moving through the vessels and how an internal organ is moving. This is why they are useful during pregnancy as they can be used to observe the structure and movement of the fetus. They are especially useful when it comes to seeing the interface between spaces that are solid and those that are filled with fluid. The field
of view depends on the shape of the probe, and the frequency of the emitted sound waves determines the depth to which they penetrate.

The uses and benefits

The technician or doctor operating the device is able to set and change the duration and frequency of the pulses.

The reason why a gel is used on a patient’s skin is to prevent any distortion of the sound waves. It is a conductive medium that forms a tight connection between the probe and the skin and reduces static.

The person using the transducer must be trained in how to operate the device and understand the human anatomy to obtain the best results. When it is done correctly, it can determine the shape, size,
and consistency of the organs and soft tissues.

The fact that transducers don’t use radiation means they don’t carry the risks associated with x-rays. They also provide a better image of soft tissue than seen on an x-ray.

We tend to associate ultrasound with monitoring of the fetus during pregnancy, but it has many other uses. It can be used to detect changes in the appearance of organs, tissues, and vessels or abnormal masses, so it is a great diagnostic tool. It can even help to provide real-time imaging during procedures such as biopsies.

A Doppler ultrasound measures movement of blood cells through the vessels and can help a physician to see blockages, blood clots and narrowing of the vessels. An echocardiogram is used to see images of the heart. A vascular transducer is normally used for carotid arteries and veins. An abdominal transducer is used in organs such as the stomach, kidney, spleen, and liver.

A cardiac transducer is typically the smallest type; followed by the vascular and then the abdominal transducer which has a larger footprint. Transvaginal transducers are long and thin with a small
head.

Two-dimensional (2-D), three-dimensional (3-D) and even four-dimensional (4-D) ultrasound is now possible, with 4-D ultrasound offering a moving image.
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Third stage

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Rad36
**Ultrasonic transducers** and **ultrasonic sensors** are devices that generate or sense ultrasound energy. They can be divided into three broad categories: transmitters, receivers and transceivers. Transmitters convert **electrical signals** into **ultrasound**, receivers convert ultrasound into electrical signals, and transceivers can both transmit and receive ultrasound.

In a similar way to **radar** and **sonar**, ultrasonic transducers are used in systems which evaluate targets by interpreting the reflected signals. For example, by measuring the time between sending a signal and receiving an echo the distance of an object can be calculated. Passive ultrasonic sensors are basically microphones that detect ultrasonic noise that is present under certain conditions.

The design of transducer can vary greatly depending on its use: those used for medical diagnostic purposes, for example the range-finding applications listed above, are generally lower power than those used for the purpose of changing the properties of the liquid medium, or targets immersed in the liquid medium, through chemical, biological or physical (e.g. erosive) effects. The latter class include ultrasonic probes and ultrasonic baths, which apply ultrasonic energy to agitate particles, clean, erode, or disrupt biological cells, in a wide range of materials;
Applications and performance[edit]

Ultrasound can be used for measuring wind speed and direction (anemometer), tank or channel fluid level, and speed through air or water. For measuring speed or direction, a device uses multiple detectors and calculates the speed from the relative distances to particulates in the air or water. To measure tank or channel liquid level, and also sea level (tide gauge), the sensor measures the distance (ranging) to the surface of the fluid. Further applications include: humidifiers, sonar, medical ultrasonography, burglar alarms, non-destructive testing and wireless charging.

Systems typically use a transducer which generates sound waves in the ultrasonic range, above 18 kHz, by turning electrical energy into sound, then upon receiving the echo turn the sound waves into electrical energy which can be measured and displayed.

This technology, as well, can detect approaching objects and track their positions[1].

Ultrasound can also be used to make point-to-point distance measurements by transmitting and receiving discrete bursts of ultrasound between transducers. This technique is known as Sonomicrometry where the transit-time of the ultrasound signal is measured electronically (ie digitally) and
converted mathematically to the distance between transducers assuming the speed of sound of the medium between the transducers is known. This method can be very precise in terms of temporal and spatial resolution because the time-of-flight measurement can be derived from tracking the same incident (received) waveform either by reference level or zero crossing. This enables the measurement resolution to far exceed the wavelength of the sound frequency generated by the transducers.

Ultrasonic transducers convert AC into ultrasound, as well as the reverse. Ultrasonics, typically refers to piezoelectric transducers or capacitive transducers. Piezoelectric crystals change size and shape when a voltage is applied; AC voltage makes them oscillate at the same frequency and produce ultrasonic sound. Capacitive transducers use electrostatic fields between a conductive diaphragm and a backing plate.

The beam pattern of a transducer can be determined by the active transducer area and shape, the ultrasound wavelength, and the sound velocity of the propagation medium. The diagrams show the sound fields of an unfocused and a focusing ultrasonic transducer in water, plainly at differing energy levels.

Since piezoelectric materials generate a voltage when force is applied to them, they can also work as ultrasonic detectors. Some systems use
separate transmitters and receivers, while others combine both functions into a single piezoelectric transceiver.

**Ultrasound Transducer Types and How to Select the Right Transducer**

To utilize the full potential of your ultrasound system, you need the right accessories. Therefore, the correct ultrasound transducer type is the key to the performance of your ultrasound.

In this blog post, we will explain the different ultrasound transducer types and determine the types of examinations you can use them for. In the end, we will offer some good points you should keep in mind when you are purchasing transducers.

**Ultrasound Transducer Types**

You can find ultrasound transducers in different shapes, sizes, and with diverse features. That is because you need different
specifications for maintaining image quality across different parts of the body.

Transducers can be either passed over the surface of the body – external transducers or can be inserted into an orifice, such as the rectum or vagina – these are internal transducers.

Any more differences?

Yes!

The ultrasound transducers differ in construction based on:

- Piezoelectric crystal arrangement
- Aperture (footprint)
- Frequency

Below we list the three most common ultrasound transducer types – linear, convex (standard or micro-convex), and phased array.

Furthermore, we have included other transducers that are available on the market and can be found in our warehouse

### Linear Transducers

*Ultrasound transducers* that produce images via **linear array** typically contain 256-512 elements, making them the largest assembly. Each element produces a scan line that makes up the ultrasound image.
Multiple adjacent elements combine to produce an ultrasound beam that is emitted at 90 degrees to the transducer head. Multiple elements (5 to 20) work as an individual unit in order to achieve a wider aperture and more useful beam shape. The ultrasound beam produced moves sequentially along the transducer one element at a time to obtain the image. Received ultrasound echoes are interpreted as corresponding to the center element of the beam.

Linear transducers produce a rectangular field of view with uniform beam density throughout. They are useful for imaging shallow structures and small parts.

So, what features are typical for the linear transducer (such as GE 9L)?
Firstly, the piezoelectric crystal arrangement is linear, the shape of the beam is rectangular (see picture below), and the near-field resolution is good.
Secondly, the footprint, frequency, and applications of the linear transducer depend on whether the product is for 2D or 3D imaging. Furthermore, the linear transducer for 2D imaging has a wide footprint and its central frequency is 2.5Mhz – 12Mhz.
You can use this transducer for various applications, for instance:

- Vascular examination
- Venipuncture, blood vessel visualization
- Breast
- Thyroid
- Tendon, arthrogenous
- Intraoperative, laparoscopy
- The thickness measurement of body fat and musculus for daily health care check and locomotive syndrome check
- Photoacoustic imaging, ultrasonic velocity change imaging

The linear transducer for 3D imaging has a wide footprint and a central frequency of 7.5Mhz – 11Mhz.

What can you use this transducer for?

- Breast
- Thyroid
- Arteria carotis of vascular application
**Convex Transducers**

The **convex ultrasound transducer** (such as GE C1-6) type is also called the curved transducer because the piezoelectric crystal arrangement is curvilinear.

Moreover, the beam shape is convex (see picture below) and the transducer is good for in-depth examinations.
Even though the image resolution decreases when the depth increases.
The footprint, frequency, and applications also depend on whether the product is for 2D or 3D imaging.
Finally, the convex transducer for 2D imaging has a wide footprint and its central frequency is 2.5MHz – 7.5MHz.
You can use it for:
- Abdominal examinations
- Transvaginal and transrectal examinations
- Diagnosis of organs

The convex transducer for 3D imaging has a wide field of view and a central frequency of 3.5MHz – 6.5MHz.
You can use it for abdominal examinations.
In addition to the convex transducers, there is a subtype called micro convex.
It has a much smaller footprint and typically, physicians would use it in neonatal and paediatrics applications.
Other Ultrasound Transducer Types
We are not done, yet. There are more ultrasound transducer types on the market. Such as:

Pencil transducers (picture below on the right), also called CW Doppler probes, are utilized to measure blood flow and speed of sound in blood.
This probe has a small footprint and uses low frequency (typically 2Mhz– 8Mhz). Furthermore, there is the **endocavitary** (picture below on the left) ultrasound transducer type. These probes provide you with the opportunity to perform internal examinations of the patient. Therefore, they are designed to fit in specific body orifices. The endocavitary transducers include **endovaginal, endorectal, and endocavity transducers**. Typically, they have small footprints and the frequency varies in the range of 3.5Mhz – 11.5Mhz. In addition, there is a **transesophageal (TEE) probe**. As well as the previously mentioned probes, it has a small footprint and is used for internal examinations. It is often employed in cardiology to obtain a better image of the heart through the oesophagus. The frequency is middle, in the range of 3Mhz – 10Mhz. Moreover, there are several probes designed for surgical use, for instance – laparoscopic probes.
ATTENUATION OF ULTRASOUND

As an ultrasound beam penetrates a medium, energy is removed from the beam by absorption, scattering, and reflection. These processes are summarized in Figure below. As with x rays, the term attenuation refers to any mechanism that removes energy from the ultrasound beam. Ultrasound is “absorbed” by the medium if part of the beam’s
Constructive and destructive interference effects characterize the echoes from non specular reflections. Because the sound is reflected in all directions, there are many opportunities for waves to travel different pathways. The wave fronts that return to the transducer may constructively or destructively interfere at random. The random interference pattern is known as “speckle."
This Waveform

This Waveform

Constructive Interference

(a)

This Waveform

This Waveform

Destructive interference

(b)
Waves can exhibit interference, which in extreme cases of constructive and destructive interference leads to complete addition (a) or complete cancellation (b) of the two waves.

Summary of interactions of ultrasound at boundaries of materials
Absorption
Reflection
Scattering
Refraction
Diffraction
Interference
Divergence

ergy is converted into other forms of energy, such as an increase in the random motion of molecules. Ultrasound is “reflected” if there is an orderly deflection of all or part of the beam. If part of an ultrasound beam changes direction in a less orderly fashion, the event is usually described as “scatter.”

The behavior of a sound beam when it encounters an obstacle depends upon the size of the obstacle compared with the wavelength of the sound. If the obstacle’s size is large compared with the wavelength of sound (and if the obstacle is relatively smooth), then the beam retains its integrity as it changes direction. Part of the sound beam may be reflected and the remainder transmitted through the obstacle as a beam of lower intensity.
If the size of the obstacle is comparable to or smaller than the wavelength of the ultrasound, the obstacle will scatter energy in various directions. Some of the ultrasound energy may return to its original source after “nonspecular” scatter, but probably not until many scatter events have occurred. In ultrasound imaging, specular reflection permits visualization of the boundaries between organs, and nonspecular reflection permits visualization of tissue parenchyma (Figure). Structures in tissue such as collagen fibers are smaller than the wavelength of ultrasound. Such small structures provide scatter that returns to the transducer through multiple pathways. The sound that returns to the transducer from such nonspecular reflectors is no longer a coherent beam. It is instead the sum of a number of component waves that produces a complex pattern of constructive and destructive interference back at the source.

This interference pattern, known as “speckle,” provides the characteristic ultrasonic appearance of complex tissue such as liver. The behavior of a sound beam as it encounters an obstacle such as an interface between structures in the medium is summarized in Figure. As illustrated in Figure 19-4, the energy remaining in the beam decreases approximately exponentially with the depth of penetration of the beam into the medium.
The reduction in energy (i.e., the decrease in ultrasound intensity) is described in decibels, as noted earlier.

Energy remaining in an ultrasound beam as a function of the depth of penetration of the beam into a medium.

**REFLECTION**

In most diagnostic applications of ultrasound, use is made of ultrasound waves reflected from interfaces between different tissues in the patient. The fraction of the impinging energy reflected from an interface depends on the difference in acoustic impedance of the media on opposite sides of the interface. In this discussion, reflection is assumed to occur at interfaces that have dimensions greater than the ultrasound wavelength. In this
case, the reflection is termed specular reflection. The acoustic impedance $Z$ of a medium is the product of the density $\rho$ of the medium and the velocity of ultrasound in the medium

$$Z = \rho c$$

Acoustic impedances of several materials are listed in the margin. For an ultrasound wave incident perpendicularly upon an interface, the fraction $\alpha R$ of the incident energy that is reflected (i.e., the reflection coefficient $\alpha R$) is

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

where $Z_1$ and $Z_2$ are the acoustic impedances of the two media. The fraction of the incident energy that is transmitted across an interface is described by the transmission coefficient $\alpha T$, where

$$\alpha_T = \frac{4Z_1Z_2}{(Z_1 + Z_2)^2}$$

Obviously $\alpha T + \alpha R = 1$
the ultrasound waves will enter the patient with little reflection at the skin surface. Similarly, strong reflections of ultrasound occur at the boundary between the chest wall and the lungs and at the millions of air–tissue interfaces within the lungs. Because of the large impedance mismatch at these interfaces, efforts to use ultrasound as a diagnostic tool for the lungs have been unrewarding. The impedance mismatch is also high between soft tissues and bone, and the use of ultrasound to identify tissue characteristics in regions behind bone has had limited success. The discussion of ultrasound reflection above assumes that the ultrasound beam strikes the reflecting interface at a right angle. In the body, ultrasound impinges upon interfaces at all angles. For any angle of incidence, the angle at which the reflected ultrasound energy leaves the interface equals the angle of incidence of the ultrasound beam; that is,

\[
\text{Angle of incidence} = \text{Angle of reflection}
\]

In a typical medical examination that uses reflected ultrasound and a transducer that both transmits and detects ultrasound, very little reflected energy will be detected if the ultrasound strikes the interface at an angle more than about 3 degrees from perpendicular. A smooth reflecting interface must be essentially perpendicular to the
ultrasound beam to permit visualization of the interface.

Ultrasound reflection at an interface, where the angle of incidence $\theta_i$ equals the angle of reflection

**REFRACTION**

As an ultrasound beam crosses an interface obliquely between two media, its direction is changed (i.e., the beam is bent). If the velocity of ultrasound is higher in the second medium, then the beam enters this medium at a more oblique (less steep) angle. This behavior of ultrasound transmitted obliquely across an interface is termed refraction. The relationship between incident and refraction angles is described by Snell’s law
For example, an ultrasound beam incident obliquely upon an interface between muscle (velocity 1580 m/sec) and fat (velocity 1475 m/sec) will enter the fat at a steeper angle.

If an ultrasound beam impinges very obliquely upon a medium in which the ultrasound velocity is higher, the beam may be refracted so that no ultrasound energy enters the medium. The incidence angle at which refraction causes no ultrasound to enter a medium is termed the critical angle $\theta_c$. For the critical angle, the angle of refraction is 90 degrees, and the sine of 90 degrees is 1. From upper equation

$$\frac{\sin \theta_i}{\sin \theta_r} = \frac{c_i}{c_r}$$

but $\sin 90^\circ = 1$
therefore

$$\theta_c = \sin^{-1}\left[\frac{c_i}{c_r}\right]$$

where $\sin^{-1}$, or arcsin, refers to the angle whose sine is $c_i / c_r$. For any particular interface, the critical angle depends only upon the velocity of ultrasound in the two media separated by the interface.

Refraction of ultrasound at an interface, where the ratio of the velocities of ultrasound in the two media is related to the sine of the angles of incidence and refraction
For an incidence angle $\theta_c$ equal to the critical angle, refraction causes the sound to be transmitted along the surface of the material. For incidence angles greater than $\theta_c$, sound transmission across the interface is prevented by refraction.
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Most of us have heard of the Doppler effect, which is the perceived change in frequency as a sound source moves toward or away from you. Since sound is a mechanical disturbance, the frequency perceived is the effective periodicity of the wavefronts. If the source is moving directly toward the observer with a velocity \( c_s \) in a medium with a speed of sound \( c_0 \) then the arriving crests appear closer together, giving the observer the acoustic illusion of a higher frequency. As illustrated in Figure 11.1, the perceived frequency depends on the direction in which the source is moving toward or away from the observer. Pierce (1989) has shown that the perceived
frequency is related to the vector dot product of the source \((c_s)\) and unit observer \((u_0)\) vectors, which differ by an angle \(\theta\), \(v_x = c_s \cdot u_0 = c_s \cos \theta\),

\[
f = f_0 + \left(\frac{f_D}{c_0}\right)c_s \cdot u_0 \tag{11.1a}
\]

and solving for the Doppler frequency \((f_D)\) in terms of the transmitted frequency \((f_0)\),

\[
f_D = \frac{f_0}{1 - (c_s/c_0) \cos \theta} \tag{11.1b}
\]

leads to a Doppler shift, correct to first order when \(c_s = c_0\),

\[
\Delta f = f_D - f_0 = f_0(c_s/c_0) \cos \theta \tag{11.1c}
\]
From this equation, the perceived frequencies for the observers in Figure 11.1 can be calculated for a 10-kHz source tone moving at a speed of 100 km/hr ($v = 27.78$ m/s).

Figure 11.1 Doppler-shifted wave frequencies from a moving source as seen by observers at different location and at the following angles relative to the directions of the source: (A) 0; (B) 90; (C) 180; (D) 270; (E) 45
in air \((c_0= 330 \text{ m/s})\). Observers B and D, at 90° to the source vector, hear no Doppler shift. Observer A detects a frequency of 10,920 Hz, while observer C (here, \(y = p\)) hears 9,220 Hz. A similar argument yields an equation for a stationary source and a moving observer with a velocity \((c_{obs})\),

\[
f = [1 + (v_{obs}/c_0) \cos \theta]f_0
\]  
(11.2)

The Doppler effect plays with our sense of time, either expanding or contracting the timescale of waves sent at an original source frequency \((f_0)\). Furthermore, it is important to bear in mind the bearing or direction of the sound relative to the observer in terms of vectors. Now consider a flying bat intercepting a flying mosquito based on the Doppler effect caused by the relative motion between them (see Figure 11.2). It is straightforward to show that if the mosquito source has a speed of \(c_s\), and the bat has a speed of \(c_{obs}\), the corresponding equation for the Doppler-shifted frequency is

\[
f = f_0\left[1 + \left(c_{obs}/c_0\right) \cos \theta\right]/\left[1 - \left(c_s/c_0\right) \cos \theta\right] = f_0\left[c_0 + c_{obs} \cos \theta\right]/\left[c_0 - c_s \cos \theta\right]
\]  
(11.3)
In other words, the flying mosquito perceives the bat signal as being Doppler shifted, and the bat hears the echo as being Doppler shifted again due to its motion. Of course, this situation is simplified greatly, as is Figure 11.1, because it is depicted twodimensionally. This description has been adequate for most medical ultrasonics, in

Figure 11.2 Bat detects insect target with ultrasound pulse echoes
which imaging is done in a plane, until the comparatively recent introduction of 3D imaging. The aero-duel between the bat and insect is played out three-dimensionally in realtime. The poor mosquito beats its wings about 200 flaps/sec, which is the annoying whine you may hear just as you are about to fall asleep on a hot night. The moth also acts as an acoustic sound source at a softer 50 flaps/sec. Enter the bat which, depending on the type, has an ultrasound range between 20 and 150 kHz (e.g., the range of the horseshoe bat is 80–100 kHz). This corresponds to an axial resolution of 2–15 mm, which is perfect for catching insects. The bat emits an encoded signal, correlates the echo response in an optimum way (shown to be close to the theoretical possible limit), adapts its transmit wave form necessary as it closes in on its target, changes its flight trajectory, and usually intercepts the insect with a resolution comparable to the size of its mouth, all in real time. Researchers are still trying to understand this amazing feat of signal processing and acrobatics and how a bat utilizes the Doppler shift between it and a fast-moving insect in 3D and while changing trajectories. Studies have shown how a bat interprets the following clues: Doppler shift (the relative speed of prey); time delay (the
distance to the target); frequency and amplitude in relation to distance (target size and type recognition); amplitude and delay reception (azimuth and elevation position); and flutter of wings (attitude and direction of insect flight). One of the key signal processing principles a bat utilizes is the repetitive interrogation of the target so that the bat can build an image of the location and speed of its prey, pulse by pulse. One of the earliest instances of pulse-echo Doppler ultrasound is in the original patent submitted by Constantin Chilowsky and Paul Langevin (1919) in 1916. Recall from Chapter 1 that their invention made underwater pulse-echo ranging technologically possible as a follow-up to earlier patents by Richardson (1913) (who also mentioned the Doppler shift but as a problem) for acoustic iceberg detection to prevent another Titanic disaster. In their patent, they mention a method to detect
Figure 11.3 Sound beam intersecting blood moving at velocity $v$ in a vessel tilted at angle $\gamma$.

relative motion between the observer and target by comparing the Doppler-shifted frequency from the target to the frequency of a stable source. The dot product results from the moving source, and moving observer cases can be applied to the simplified situation of a transducer sensing the flow of blood in a vessel flowing with velocity and direction ($v_t$) at an angle ($\gamma$) to the vessel, as depicted in Figure 11.3. In this case, the transducer is infinitely wide and the intervening tissues have negligible effect. The blood velocity is much smaller than the speed of sound in the intervening medium ($c_0$). The
signal as seen from an observer riding the moving blood appears to be Doppler shifted,

$$\omega_T = \omega_2 + c_D \cdot k_i = \omega_0 + c_D \cdot \left( \frac{\omega_i}{c_0} \right) n_i \quad (11.4a)$$

where $\omega_T$ is the shifted angular frequency, $\omega_2$ is the angular frequency seen by the scattering object from a moving coordinate system, $\omega_i$ is the incident angular frequency, $c_D$ is the Doppler velocity, and $n_i$ is in the direction of the incident $k$ vector along the beam. The returning scattered signal along unit vector $n_{sc}$ appears to be from a moving source and is Doppler shifted,

$$\omega_R = \omega_2 + c_D \cdot k_{sc} = \omega_2 + c_D \cdot \left( \frac{\omega_{sc}}{c_0} \right) n_{sc} \quad (11.4b)$$

where $\omega_R$ is the shifted angular frequency, $\omega_{sc}$ is the scattered frequency Doppler, and $n_{sc}$ is in the direction of the scattered $k$ vector back toward the
transducer. For a coincident transmitter and receiver, the overall Doppler shift can be found by subtracting the the \( \omega_R \) from \( \omega_T \) and letting \( \omega_T \) 0 to first order,

\[
\omega_R - \omega_T = \omega_0 \left( \frac{c_D}{c_0} \right) [1 + \cos(\theta) - \cos(\pi - \theta)] = \omega_0 \left( \frac{c_D}{c_0} \right) [2 \cos \theta]
\]  

(11.4c)

the \( \omega_R \) from \( \omega_T \) and letting \( \omega_T \) 0 to first order,

\[
f_D = \Delta f = f_R - f_T = \left| \frac{2v}{c_0} \cos \theta \right| f_0
\]

(11.4d)

Before looking at ways that this Doppler shift can be implemented in instrumentation, it is worth understanding more about the properties of blood and how it interacts with sound.
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Competent use of Doppler ultrasound techniques requires an understanding of three components: the capacity and limitations of Doppler ultrasound, the parameters that contribute to a flow assessment and the features of blood flow in arteries and veins. Ultrasound images of flow are obtained by measuring moving fluids. In ultrasound scanners, a series of pulses is transmitted to detect the movement of blood; echoes from moving scatterers determine slight differences in the time for return of the signal to the receiver. These differences can be measured as direct time differences or, more often, in terms of a phase shift from which the Doppler frequency is obtained (Fig. 2.127). They are then processed to produce a colour flow display or a Doppler sonogram.
Fig. 2.127 shows the Doppler transducer scanning at an angle $\theta$ to a blood vessel, in which blood flows at a velocity of $u \text{ m/s}$. The ultrasound waves are emitted by the transducer at a frequency $f_0$ and are directed back to the transducer by moving reflectors in the blood (red blood cells) at a different frequency $f_r$. The difference between the transmitted and received frequencies, $\Delta f$, is related to the velocity of the flowing blood, $u$, and the speed of sound in tissue, $v$, according to the equation:
There has to be motion in the direction of the beam to obtain Doppler signals; this does not happen if the flow is perpendicular to the beam.

The magnitude of the Doppler signal depends on:

- blood velocity;
- ultrasound frequency: the higher the ultrasound frequency, the higher the Doppler frequency. As in B-mode, lower ultrasound frequencies have better penetration; the choice of frequency is therefore a compromise between better sensitivity and better penetration;
- the angle of insonation: the Doppler frequency increases as the angle between the beam and the direction of flow becomes smaller.

\[ \Delta f = f_o - f_r - \frac{2 f_o u \cos \theta}{v} \]
The two main types of Doppler systems in common use today are continuous wave and pulsed wave. They differ in transducer design, operating features, signal processing procedures, and the types of information provided.

Doppler practice

Continuous-wave Doppler requires continuous generation of ultrasound waves with continuous ultrasound reception. A two-crystal transducer has this dual function, with one crystal for each function.

The main advantage of continuous-wave Doppler is for measuring high blood velocities accurately;

Its main disadvantage is the lack of selectivity and depth discrimination. As continuous-wave Doppler is constantly transmitting and receiving from two different transducer heads (crystals), there is no provision for imaging or range gating to allow selective placing of a given Doppler sample volume in space.
Pulsed-wave Doppler systems have a transducer that alternates transmission and reception of ultrasound in a way similar to the M-mode transducer. The main advantage of pulsed Doppler is that Doppler shift data are produced selectively from a small segment along the ultrasound beam, referred to as the sample volume, the location of which is controlled by the operator. An ultrasound pulse is transmitted into tissues and travels for a given time (time X) until it is reflected back by a moving red cell. It then returns to the transducer over the same interval but at a shifted frequency. The total transit time to and from the area is 2X. This process is repeated alternately through many transmit–receive cycles each second. The range gating therefore depends on a timing mechanism that samples the returning Doppler shift data from only a given region. All other returning ultrasound information is essentially ignored. The main disadvantage of pulsed-wave Doppler is that high blood flow velocities cannot be measured accurately.
Aliasing effect:

Pulsed-wave systems have a fundamental limitation. When pulses are transmitted at a given sampling frequency (pulse repetition frequency), the maximum Doppler frequency that can be measured is half of that frequency. Therefore, if the Doppler frequency is greater than half the pulse repetition frequency, the Doppler signal is ambiguous. This condition is known as aliasing. The interval between sampling pulses must be sufficient for a pulse to go away and come back to the transducer. If a second pulse is sent before the first is received, the receiver cannot correctly discriminate between the two. The deeper the sample volume, the longer the pulse’s journey, so that the pulse repetition frequency must be reduced for unambiguous ranging. The result is that the maximum measurable Doppler frequency decreases with depth.
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Ultrasound Doppler modes:

**Colour flow imaging and spectral Doppler:** Colour flow imaging produces a picture of a blood vessel by converting the Doppler data into colours, which are overlaid onto the B-mode image of the blood vessel and represent the speed and direction of blood flow through the vessel. It is useful for identifying the vessels under examination, verifying the presence and direction of flow and finding the correct angle.

**Power Doppler** is a newer ultrasound technique that is up to five times more sensitive in detecting low blood flow than standard colour Doppler. The magnitude of the flow output rather than the Doppler frequency signal is shown. Power Doppler can obtain some images that are difficult or impossible to obtain with standard colour Doppler.
Spectral Doppler (pulsed-wave Doppler)

is used to provide a sonogram of a vessel in order to measure the distribution of flow velocities in the sample volume. The Doppler signal is processed in a Fourier spectrum analyser. The amplitudes of the resulting spectra are encoded as brightness, and these are plotted as a function of time (horizontal axis) and frequency shift (vertical axis) to give a two-dimensional spectral display. With this technique, a range of blood velocities in a sample volume will produce a corresponding range of frequency shifts on the spectral display.
To obtain proper images and measures, the operator should:

- identify the vessel (possibly by colour flow imaging);
- adjust the gain so that the sonogram is clear and free of noise;
- place the Doppler cursor on the vessel to be investigated (sample volume) and set the correct size;
- obtain a proper angle of insonation (60° or less);
- adjust the pulse repetition frequency to suit flow conditions and avoid aliasing
Flow waveform analysis:

Doppler waveform analysis is often used for diagnosis in the clinical assessment of disease. The complex shapes of Doppler waveforms can be described by relatively simple indices, which have been used to evaluate fetal health and organ blood flow. Common indices are the pulsatility index (PI), resistance index (RI) and the ratio of systolic to diastolic (S/D or A/B).

\[
PI = \frac{f_{\text{max}} - f_{\text{min}}}{f_{\text{mean}}}
\]

\[
RI = \frac{f_{\text{max}} - f_{\text{min}}}{f_{\text{max}}}
\]

\[
S/D = \frac{f_{\text{max}}}{f_{\text{min}}}
\]
where $f_{\text{max}}$ is the maximum systolic frequency, $f_{\text{min}}$ is the minimum diastolic frequency, and $f_{\text{mean}}$ is the time-averaged frequency (Fig. 2.128). An advantage of these waveform indices is that they consist of ratios of Doppler shift frequencies and are thus independent of transmission frequency and Doppler angle. Generally, low- and high-pulsatility waveforms occur in low- and high-resistance vascular beds, respectively. In addition to these indices, the flow waveform can be described or categorized by the presence or absence of a particular feature, for example the absence of end-diastolic flow and the presence of a post-systolic notch.
Fig. 2.128. Doppler waveform indices (for abbreviations, see text)
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A-Mode

In the A-mode presentation of ultrasound images, echoes returning from the body are displayed as signals on an oscilloscope. The oscilloscope presents a graph of voltage representing echo amplitude (hence the term “A-mode”) on the ordinate, or $y$-axis, as a function of time on the abscissa, or $x$-axis. With the assumption of a constant speed of sound, time on the $x$-axis can be presented as distance from the ultrasound transducers (Margin Figure 21-1).
A-mode reveals the location of echo-producing structures only in the direction of the ultrasound beam. It has been used in the past to localize echo-producing interfaces such as midline structures in the brain (echoencephalography) and structures to be imaged in B-mode. A-mode displays are not found on most imaging systems used today. The concept of A-mode is however, useful in explaining how pixels are obtained from scan lines in B-mode imaging.
MARGIN FIGURE 21-1
A-mode (amplitude mode) of ultrasound display. An oscilloscope display records the amplitude of echoes as a function of time or depth. Points A, B, and C in the patient appear as peaks A, B, and C in the A-mode display.
B-Mode

In B-mode presentation of pulse echo images (Margin Figure 21-2) the location of echo-producing interfaces is displayed in two dimensions (x and y) on a video screen. The amplitude of each echo is represented by the brightness value at the xy location. High-amplitude echoes can be presented as either high brightness or low brighten to provide either “white-on-black” or “black-on-white” presentations. Most images are viewed as white on black, so regions in the patient that are more echogenic correspond to regions in the image that are brighter (hence B for “brightness” mode).
B-mode images may be displayed as either “static” or “real-time” images. In static imaging the image is compiled as the sound beam is scanned across the patient, and the image presents a “snapshot” averaged over the time required to sweep the sound beam. In real-time imaging, the image is also built up as the sound beam scans across the patient, but the scanning is performed automatically and quickly, and one image follows another in quick succession. At image frequencies greater than approximately 24 per second, the motion of moving structures seems continuous, even though it may appear that the images are flickering (i.e., being flashed on and off).
Images that are refreshed at frequencies greater than approximately 48 per second are free of flicker. Real-time B-mode images are useful in the display of moving structures such as heart valves. They also permit the technologist to scan through the anatomy to locate structures of interest. Certain applications such as sequential slice imaging of organs are performed better with static imaging.
MARGIN FIGURE 21-2
B-mode, or brightness mode, ultrasound *display*. The amplitude of reflected signals is displayed as brightness at points of an image defined by their *x*- and *y*-coordinates.
M-Mode

The M-mode presentation of ultrasound images is designed specifically to depict moving structures. In an M-mode display, the position of each echo-producing interface is presented as a function of time. The most frequent application of M-mode scanning is echocardiography, where the motion of various interfaces in the heart is depicted graphically on a cathode-ray tube (CRT) display or chart recording. In a typical M-mode display (Margin Figure 21-3), the depths of the structures of interest are portrayed as a series of dots in the vertical direction on the CRT, with the position of the transducer represented by the top of the display.
With the transducer in a fixed position, a sweep voltage is applied to the CRT deflection plates to cause the dots to sweep at a controlled rate across the CRT screen. For stationary structures, the dots form horizontal lines in the image. Structures that move in a direction parallel to the ultrasound beam produce vertical fluctuations in the horizontal trace to reveal their motion. The image may be displayed on a short-persistence CRT or storage scope and may be recorded on film or a chart recorder. Modern systems digitize the information and display it on a digital monitor.
MARGIN FIGURE 21-3

A typical M-mode echocardiographic tracing
Magnetic Resonance Imaging (MRI)

- Introduction
- The Components
- The Technology
- Physics behind MR

Introduction

What is MRI?

- Magnetic resonance imaging (MRI) is a spectroscopic imaging technique used in medical settings to produce images of the inside of the human body.

- MRI is based on the principles of nuclear magnetic resonance (NMR), which is a spectroscopic technique used to obtain microscopic chemical and physical data about molecules.

- In 1977 the first MRI exam was performed on a human being. It took 5 hours to produce one image.
Introduction

- How Does it Work?
  - The magnetic resonance imaging is accomplished through the absorption and emission of energy of the radio frequency (RF) range of the electromagnetic spectrum.
Why MRI?

- Utilizes non ionizing radiation. (unlike x-rays).
- Ability to image in any plane. (unlike CT scans).
- Very low incidents of side effects.
- Ability to diagnose, visualize, and evaluate various illnesses.

The only better way to see the insides of your body is to cut you open!
The Components:

- A magnet which produces a very powerful uniform magnetic field.
- Gradient Magnets which are much lower in strength.
- Equipment to transmit radio frequency (RF).
- A very powerful computer system, which translates the signals transmitted by the coils.
The most important component of the MRI scanner is the magnet:

- The magnets currently used in scanners today are in the .5-tesla to 2.0-tesla range (5,000 to 20,000-gauss).
  
  Higher values are used for research.

- Earth magnetic field: 0.5-gauss
The Magnet (cont.)

- There are three types of magnets used in MRI systems:
  - Resistive magnets
  - Permanent magnets
  - Super conducting magnets (the most commonly used type in MRI scanners).

- In addition to the main magnet, the MRI machine also contains three gradient magnets. These magnets have a much lower magnetic field and are used to create a variable field.
Spin: The atoms that compose the human body have a property known as spin (a fundamental property of all atoms in nature like mass or charge).

Spin can be thought of as a small magnetic field and can be given a + or – sign and a mathematical value of multiples of ½.

Components of an atom such as protons, electrons and neutrons all have spin.
Spin (cont.):

- Protons and neutron spins are known as nuclear spins.
- An unpaired component has a spin of $\frac{1}{2}$ and two particles with opposite spins cancel one another.
- In NMR it is the unpaired nuclear spins that produce a signal in a magnetic field.
Human body is mainly composed of fat and water, which makes the human body composed of about 63% hydrogen.

Why Are Protons Important to MRI?
- positively charged
- spin about a central axis
- a moving (spinning) charge creates a magnetic field.
- the straight arrow (vector) indicates the direction of the magnetic field.
The Technology (cont.)

- When placed in a large magnetic field, hydrogen atoms have a strong tendency to align in the direction of the magnetic field.
- Inside the bore of the scanner, the magnetic field runs down the center of the tube in which the patient is placed, so the hydrogen protons will line up in either the direction of the feet or the head.
- The majority will cancel each other, but the net number of protons is sufficient to produce an image.
The MRI machine applies radio frequency (RF) pulse that is specific to hydrogen.

- The RF pulses are applied through a coil that is specific to the part of the body being scanned.
The Technology (Cont.)

Resonance (cont.)

The gradient magnets are rapidly turned on and off which alters the main magnetic field.

- The pulse directed to a specific area of the body causes the protons to absorb energy and spin in different direction, which is known as resonance.

*Frequency (Hz) of energy absorption depends on strength of external magnetic field.*
The Technology (cont.)

Larmor Equation

\[ \omega_0 = \gamma \beta_0 / 2\pi \]

For hydrogen at 1.5T:

\[ \gamma = 2.675 \times 10^8 \frac{1}{s \cdot T} \]
\[ \beta_0 = 1.5T \]
\[ \omega_0 = 63.864 MHz \]

- the resonance frequency, \( \omega_0 \), is referred to as the Larmor frequency
- this frequency is needed to excite transverse magnetization (precession)
The Technology (cont.)

- Imaging:
  - When the RF pulse is turned off the hydrogen protons slowly return to their natural alignment within the magnetic field and release their excess stored energy. This is known as relaxation. -> two time-scales (see later)

- What happens to the released energy?
  - Released as heat
  - OR
  - Exchanged and absorbed by other protons
  - OR
  - Released as Radio Waves.
Measuring the MR Signal:

- The moving proton vector induces a signal in the RF antenna.

- The signal is picked up by a coil and sent to the computer system. The received signal is sinusoidal in nature.

- The computer receives mathematical data, which is converted through the use of a Fourier transform into an image.
Measuring the MR Signal

original long. alignment

RF signal from precessing protons

transverse precession

RF antenna
Physics of MRI

It is an interplay of

- Magnetism
- Resonance
**Fig: 1. A)** The top spinning in the earth's gravity. The gravity tries to pull it down but it stays upright due to its fast rotation. **B)** A charge spinning in the magnetic field Bo.
A) The protons spinning in the nature, without an external strong field. The directions of spins are random and cancel out each other. The net magnetization is nearly 0.

B) In the presence of a large external magnetic field $B_0$ the spins align themselves either against (high energy state) or along (low energy state). There is a slight abundance of spins aligned in the low energy state.
Longitudinal magnetization

more spins with lower energy,
i.e. parallel to external field
A) The compass needle (a small magnet) aligns itself with a N/S-S/N direction when placed in a large magnetic field.

B) When another strong magnet is brought near the aligned compass needle the magnetic fields of all three magnets interact in such a way that the mobile, weakest magnet (the compass needle) realigns itself away from its original orientation.

C) When the perturbing magnetic field is removed suddenly the compass needle magnet realigns itself with the large external magnet field, but before realigning, it wobbles around the point of stability and gradually comes to rest.
Fig: 4. The spin of a proton can be represented by a vector $B$ with a direction and magnitude. Its relation to the direction of the external magnetic field $B_0$ is represented by an angle.
A) The spin of a proton aligned to Bo in the Z-axis.
B) An external perturbing magnetic field, B1, is applied which knocks the vector out of its axis, which now is aligned at a new angle with respect to Bo.
C) As the perturbing field B1 is removed the vector gradually starts returning back to its original state and
D) begins to wobble
A) The falling water rotates a wheel to which a magnet is attached. When this magnet rotates it induces an alternating current in a coil of wire which can be detected.

B) A magnetic field (spin of a proton) rotating near a coil of MR antenna induces a similar current in the loop which can be detected.
Resonance

The gradient coils.
- **A)** the body placed in the core of the magnet with $B_0$ aligned to its long axis.
- **B)** the gradient coil oriented in the Z-axis (along the long axis of the body) which gradually and linearly increases from left to right.
- **C)** At the center of the gradient field, the frequency is equal to that of $B_0$, but at a distance $\Delta x$ the field changes by a factor of $\Delta B_0$. 
Resonance

- The resonance equation shows that the resonance frequency \( \nu \) of a spin is proportional to the magnetic field, \( B_0 \), it is experiencing.

\[
\nu = \gamma B_0
\]

- Where \( \gamma \) is the gyromagnetic ratio. [*the ratio of the magnetic moment of a spinning charged particle to its angular momentum*]
Two time-scales: T1 and T2

- T1 is the spin-lattice relaxation time-scale for the longitudinal magnetization to come back to its initial value
  \[ M_z(t) = M_0 \cdot (1 - c \cdot e^{-\frac{t}{T_1}}) \]

- T2 is the spin-spin relaxation time for the transverse magnetization to decrease to zero
  \[ M_T(t) = M_T(0) \cdot e^{-\frac{t}{T_2}} \]

- T1-weighted imaging: higher spatial resolution
- T2-weighted imaging: higher tissue contrast
- typically both time-scales are used
Recap: What Does the Image Represent?

- For every unit volume of tissue, there is a number of cells, these cells contain water molecules, each water molecule contain one oxygen and two hydrogen atoms.

- Each hydrogen atom contains one proton in its nucleus. Different tissues thus produce different images based on the amount of their hydrogen atoms producing a signal.
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