Module 4

Ultrasound Imaging

VASCPROG 560

Vascular Imaging Techniques

CIHR Strategic Training Program in Vascular Research
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Instead of a course textbook, all the modules contain links to excellent information that can be found on the internet. It is important that you visit these links to get more background on the topics. These also may be printed out to read in more detail later, or to be saved for future reference.

If you have any difficulty in accessing any of the links within these modules please send an email to jwilliams@robarts.ca. Sometimes the sources of the links change and adjustments will be made to correct this.

When you have finished the module, please go to the Module 4 Quiz under the Quizzes icon on the Course Home Page.
Information in this module was based on lecture notes given by Robarts Scientists as part of the coursework in the Department of Biomedical Engineering at the University of Western Ontario, and from material from the following sources:


Jackie Williams wrote some of the module content and organized it for WebCT.

I would like to thank Dr. Donal B. Downey, Interventional Radiologist at London Health Sciences Centre, London, Ontario, who provided input on this Module.

Thanks also to Dr. James Lacefield, who allowed use of his lecture notes from his teaching module, “Ultrasound Imaging and Bioeffects”, which form part of the “Physics for Radiology Residents Review Course.”
Objectives

After completing this module, the student should be able to:

- Define ultrasound, reflection, refraction, and absorption, and describe how ultrasonic waves are affected by propagation through a medium.
- Calculate the wavelength of diagnostic ultrasound from its frequency and its speed through tissue.
- Describe how the speed of sound through a material is related to characteristics of the material.
- Define specular reflection and calculate the reflection coefficient at the interface of two materials.
- Calculate the angle of incidence or angle of refraction if given one of the two and the speed of sound for the incidence and refractive materials.
- Understand how different tissues have different acoustic impedances, and how this affects the amount of reflection.
- Define the attenuation coefficient for ultrasonic waves traveling through biological tissue.
- Understand how piezoelectric crystals are used to produce and detect ultrasound.
- Draw a typical ultrasound transducer and briefly describe its components.
- Define Fresnel zone and Fraunhofer zone.
- Briefly describe three modes for displaying diagnostic ultrasound (A-mode, B-mode, M-mode).
- Describe a phased array transducer for real-time scanning.
INTRODUCTION

Ultrasound (also known as sonography or ultrasonography) is a medical diagnostic technique which allows the visualization and examination of different parts of the human anatomy using high frequency sound waves, which are emitted from a probe and directed into the body.

All the different diagnostic ultrasound techniques involve the detection and display of acoustic energy reflected off different tissues within the body. Different body structures have different properties that scatter and reflect sound energy in predictable ways, making it possible to recognize these structures in the two-dimensional, gray-scale images produced by ultrasound scanners.

However, ultrasound techniques can also be used to detect and display flow parameters, making it very useful in vascular imaging, and they can also be displayed in 3-D.

There are many variables involved in producing, detecting and processing ultrasound data, which are, for the most part, under the control of the operator. Of all the different imaging techniques, ultrasound is the most influenced by the skill and experience of the operator, both in acquiring and interpreting the images.

This module will explain the basic physics of how sound waves can produce images of the human body, and the different variables that govern the quality of the ultrasound image. It will describe the advantages and problems of working with ultrasound and the different techniques that are used in vascular imaging.
Diagnostic ultrasound offers several advantages over other imaging modalities, the most important being that it does not use ionizing radiation. This makes it the safest modality and is why it is the mainstay of obstetrical imaging.

Besides its safety, the next most important advantage is its ability to image in real time, making it simple to perform live active and passive range of motion studies. The cost of an ultrasound scanner is less than 10% than an MRI scanner.

<table>
<thead>
<tr>
<th>Advantages of Ultrasonography</th>
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<tbody>
<tr>
<td>Uses no ionizing radiation</td>
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<tr>
<td>Safe in pregnancy</td>
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<tr>
<td>Has no known side effects</td>
</tr>
<tr>
<td>Inexpensive</td>
</tr>
<tr>
<td>Portable</td>
</tr>
<tr>
<td>Minimal preparation of patients</td>
</tr>
<tr>
<td>Painless</td>
</tr>
<tr>
<td>Non-invasive</td>
</tr>
<tr>
<td>Gives direct vision for biopsies</td>
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Some of the problems with ultrasound imaging are that diagnostic images sometimes cannot be obtained because of a patient’s size, or because the ultrasound beam cannot traverse air-filled or bony regions. In such cases, cross-sectional imaging with computed tomography or MRI may be used instead.
The following brief information is from an excellent website called, "A Short History of the Development of Ultrasound in Obstetrics and Gynecology", by Dr. Joseph Woo. For those who are interested in a very readable account of the history of ultrasound, the website can be found at: History of Ultrasound in Obstetrics and Gynecology

Early attempts to measure the speed of sound were made in 1822 by Daniel Colladen, a Swiss physicist, using an underwater bell. In 1877 Lord Rayleigh published the treatise "The Theory of Sound" which described the fundamental physics of sound vibrations (waves), transmission and refraction.

The important discovery that made modern ultrasound imaging possible was made in 1889 by Pierre Curie and his brother Jacques Curie in France, when they discovered the piezo-electric effect of certain crystals, observing, “that an electric potential would be produced when mechanical pressure was exerted on a quartz crystal such as the Rochelle salt (sodium potassium tartrate tetrahydrate) and conversely, the application of an electric charge would produce deformation in the crystal and causing it to vibrate. It was then possible for the generation and reception of 'ultra'-sounds that are in the frequency range of millions of cycles per second (megahertz), which could be employed in echo sounding devices."

The two World Wars were the catalyst for a great deal of research on underwater echo sounding devices. The first working sonar (Sound Navigation and Ranging) system was designed and built in the United States by a Canadian called Reginald A. Fessenden in 1914.
Brief History of Ultrasound in Medicine

Ultrasonics in medicine were first used in therapy rather than diagnosis, utilizing it's heating and disruptive effects on animal tissues.

By the 1940s, ultrasound had become a neuro-surgical tool. William Fry at the University of Illinois and Russell Meyers at the University of Iowa performed craniotomies using ultrasound to ablate parts of the basal ganglia in patients with Parkinsonism.

Ultrasound began to be used as a diagnostic tool in medicine in 1942 by Karl Theodore Dussik, a neurologist/psychiatrist at the University of Vienna. He and his brother Friederich, a physicist, measured the transmission of ultrasound beams through the skull, employing a transducer on either side. Although this was not very successful it paved the way for further research.

Throughout the 1940's and 50's, huge strides were made in technical developments in ultrasound, including better piezoelectric crystals and high input impedance amplifiers. In 1951 Howry, Holmes, Bliss, and Posakony produced the Immersion tank ultrasound system, the first 2-dimensional B-mode (or PPI, plan position indication mode) linear compound scanner. In 1954 they produced the motorized Somascope, a compound circumferential scanner. The sonographic images were referred to as somagrams. The innovative apparatus was reported in the Medicine section of LIFE Magazine® in 1954.

The Pan Scanner fabricated by the Holmes, Howry, Posakony and Cushman team in 1957 was a real breakthrough and landmark invention in the history of B-mode ultrasonography. Since then ultrasound technology has continued, and is still continuing its steady progress with ever increasing medical applications.
Sound Waves and Ultrasound Waves

- Sound waves and their properties are the basis for understanding ultrasound imaging.
- Sound waves consist of a mechanical disturbance of a medium, which can be a gas, a solid, or a liquid.
- The acoustic energy causes a physical displacement of the medium, forming a pressure wave that produces alternating compression and rarefaction.
- Changes in pressure over time define the basic units of measurement for sound.
- The following page shows the sinusoidal wave of the changes in pressure of sound travelling through a particular medium. The disturbance passes through the medium at a fixed speed and causes vibration. The rate at which the particles of the medium vibrate is the frequency, measured in cycles per second or hertz (Hz).
- If you remember from elementary physics, frequency is the pitch of a sound - the higher the frequency, the higher the pitch. Frequency is measured in hertz. In ultrasound it is often referred to in kilohertz (kHz) or megahertz (MHz). The relationship is:

\[
1 \text{ Megahertz} = 1000 \text{ Kilohertz} = 1,000,000 \text{ Hertz}
\]

- The frequency range of sound audible to the human ear is approximately 60 to 20,000 Hz.
- Beyond about 20kHz sound is known as **ULTRASOUND**. Ultrasonic frequencies well above 15,000 Hz can be detected by such mammals as bats. Infrasound (low-frequency sound) can be detected by some animals and birds. Pigeons can detect sounds as low as 0.1 Hz while elephants communicate using sounds as low as 1 Hz. Diagnostic imaging uses much higher frequencies, with the usual range being between 1.5 and 15 MHz.
Sound travels through materials under the influence of sound pressure. Because molecules or atoms of a solid are bound elastically to one another, the excess pressure results in a wave propagating through the solid. A sound wave is a series of compressions and rarefactions. The combination of one compression and one rarefaction represents one cycle. The distance between the onset (peak compression) of one cycle and the next is the wavelength. The time (T) to complete one cycle is called the Period (the time between oscillations, or 1/f).

Frequency (f) and Period are inversely related. If the period, T, is expressed in seconds, then the frequency in Hz is:

\[ f = \frac{1}{T} \]
Generation of Ultrasound

Sound propagates as longitudinal waves. A longitudinal wave is produced when a vibrator, such as a piezoelectric crystal in an ultrasound transducer, transmits its back and forth oscillation into a medium. The particles of the medium are made to oscillate in the direction of the wave propagation, but are otherwise stationary. The wave propagates as bands of compression and rarefaction. One wavelength ($\lambda$) is the distance between two bands of compression, or rarefaction.
The Propagation and Velocity of Sound

When used to image the body, ultrasound techniques use brief pulses of acoustic energy that are propagated through the tissues. The speed of the pressure wave varies according to the physical properties of the different tissues, according to their resistance to compression. The amount of resistance depends on the density and stiffness of the medium (see table on following page).

\[ V = f \lambda \]

- \( V \) (or \( c \)) = velocity (units: m/sec)
- \( f \) = frequency (units: cycles/sec or Hz)
- \( \lambda \) = wavelength (units: m or mm)

**NOTE 1:** 1 MHz = \( 10^6 \) cycles/second

**NOTE 2:** Frequency is determined by the source, i.e. the transducer/ultrasound machine.
### Velocity of Sound in Various Materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Velocity (m/sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air (quoted at 20°C)</td>
<td>330</td>
</tr>
<tr>
<td>Fat (quoted at 37°C)</td>
<td>1460</td>
</tr>
<tr>
<td>Mercury</td>
<td>1450</td>
</tr>
<tr>
<td>Castor Oil</td>
<td>1500</td>
</tr>
<tr>
<td>Water (at 50°C)</td>
<td>1540</td>
</tr>
<tr>
<td>“HUMAN SOFT TISSUE”</td>
<td>1540</td>
</tr>
<tr>
<td>Brain</td>
<td>1541</td>
</tr>
<tr>
<td>Liver (quoted at 37°C)</td>
<td>1555</td>
</tr>
<tr>
<td>Kidney</td>
<td>1561</td>
</tr>
<tr>
<td>Blood</td>
<td>1570</td>
</tr>
<tr>
<td>Muscles (quoted at 37°C)</td>
<td>1600</td>
</tr>
<tr>
<td>Lens of eye</td>
<td>1620</td>
</tr>
<tr>
<td>Skull (quoted at 37°C)</td>
<td>4080</td>
</tr>
<tr>
<td>Brass</td>
<td>4490</td>
</tr>
<tr>
<td>Aluminum</td>
<td>6400</td>
</tr>
</tbody>
</table>

- Diagnostic ultrasound machines assume an average sound speed of 1540 m/s.

- Velocity is inversely proportional to compressibility. The less compressible a medium, the greater the velocity.

- Sound travels more quickly in solids and more slowly in gases.

- The speed of sound in a vacuum is zero.
The wavelength of sound decreases as its frequency (MHz) increases. This is because velocity is constant, so increases in frequency result in decreases in wavelength and vice versa (V = fλ).

For ultrasound purposes the velocity assumed is the average of these measurements, which is 1540 m/second. As a few normal tissues differ significantly from this average value, their display may be subject to measurement errors or artifacts.
The behaviour of a wave incident on a plane boundary separating two typical media which differ in characteristic impedance.
Refraction

When sound passes from adjacent tissues having different acoustic propagation velocities then this can cause a change in direction of the sound wave. This change in direction is called refraction and is governed by

**Snell’s Law:**

$$\sin \theta_i = \sin \theta_r$$

$$\sin \theta_t = \sin \theta_i \left[ \frac{C_2}{C_1} \right]$$

Where: $\theta_i$ is the angle of incidence;
- $\theta_r$ is the angle of reflection;
- $\theta_t$ is the angle of transmission, and
- $C_1$ and $C_2$ are the velocities of sound in media 1 and 2 respectively

(See diagram on previous page)

Refraction is one of the causes of misregistration in an ultrasound image. Ultrasound scanners assume that returning echoes are coming from a fixed line of sight from the transducer. If the sound wave has been refracted, the echo in the ultrasound image may be coming from a different location than that shown in the display. This can be avoided by making the scan angle as perpendicular to the interface as possible.
1. Reflection

a. Specular - If the acoustic interface is large and smooth, it reflects sound much as a mirror reflects light. Such interfaces are called specular reflectors and examples of these in the body are the diaphragm and the urine-filled bladder.

b. Scattering - from small targets.

Specular echoes originate from relatively large, strongly reflective, regularly shaped objects with smooth surfaces and are relatively intense and angle dependent. However, some echoes in the body come from smaller interfaces within solid organs, with structures that are much smaller than the wavelength of the incident sound. The echoes from these interfaces are scattered in all directions.

2. Attenuation

a. Absorption – tissue is viscoelastic and viscous reactions convert mechanical to thermal energy.

b. Redirection – sound energy is redirected outside the imaging beam and gets lost to the imaging system.
Reflection and Transmission

The amount of ultrasound reflected or refracted depends on the angle at which the ultrasonic beam hits the interface between different media. As the angle of incidence approaches 90° (perpendicular to the interface), a higher percentage of the ultrasound is reflected.

Since ultrasound scanners only detect reflections that return to the transducer, the display of specular interfaces depends heavily on the angle of insonation. Specular reflectors (e.g., the diaphragm) only return echoes to the transducer if the sound beam is perpendicular to the interface. If not, the sound beam will be transmitted away from the transducer, and the echo will not be detected.
Diagnostic ultrasound scanners rely on reflected sound or echoes from a reflective interface. If the different body tissues were completely homogeneous then there would be no reflective interfaces and the body would appear anechoic (without echoes). Where two different tissues are juxtaposed, for example, the liver and the diaphragm, this provides an acoustic interface. Different interfaces are responsible for the reflection of varying amounts of the incident sound wave.

The amount of the reflection (or backscatter) is determined by the difference in the acoustic impedances of the tissues at the interface. Acoustic Impedance, Z, is determined by multiplying the density (\( \rho \)) of the medium propagating the sound and the propagation velocity V of sound in that medium.

\[
Z = \rho V
\]

Z (units: The Rayl, named after Lord Rayleigh, measured in kg/s m\(^2\))

\( \rho \equiv \text{density (units: kg/m}^3\))

V \equiv \text{velocity (units: m/sec)}

Interfaces with large acoustic impedance differences, such as bone and air, reflect almost all the incident energy, but those with smaller differences, such as muscle and fat, reflect only part of the incident energy, allowing the remainder to continue on the same path.
Attenuation – a weakening of the acoustic energy as it moves through a medium and transfers some of that energy as heat. This attenuation of sound energy is clinically important because it affects the depth of penetration into the tissues. It governs transducer selection and various operator-controlled instrument settings, such as time gain compensation (described in more detail later in the module), power output attenuation, and system gain levels.

As sound passes through tissue it loses energy and the pressure waves decrease in amplitude as they travel further from their source. Some of the energy is transferred to the tissues producing heat (absorption) and some of the acoustic energy is lost through reflection and scattering.

Attenuation is therefore the result of the combined effects of absorption, scattering, and reflection. Attenuation depends on the frequency of the sound wave as well as the nature of the attenuating medium. High frequencies are attenuated more rapidly than lower frequencies and the transducer frequency is a major determinant of the useful depth from which information can be obtained with ultrasound.

The measurement of acoustic power is expressed in watts (w) or milliwatts (mW) and describes the amount of acoustic energy produced in a unit of time. Scattering and absorption are both frequency and depth dependent, so attenuation is typically expressed in units of decibels/cm/ MHz.
<table>
<thead>
<tr>
<th>Tissue</th>
<th>Attenuation Coefficient (α)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>12.00 dB/cm/MHz</td>
</tr>
<tr>
<td>Bone</td>
<td>5.00 dB/cm/MHz</td>
</tr>
<tr>
<td>Muscle (transverse)</td>
<td>3.30 dB/cm/MHz</td>
</tr>
<tr>
<td>Kidney</td>
<td>1.00 dB/cm/MHz</td>
</tr>
<tr>
<td>Liver</td>
<td>0.94 dB/cm/MHz</td>
</tr>
<tr>
<td>Soft Tissue (avg)</td>
<td>0.70 dB/cm/MHz</td>
</tr>
<tr>
<td>Fat</td>
<td>0.63 dB/cm/MHz</td>
</tr>
<tr>
<td>Blood</td>
<td>0.18 dB/cm/MHz</td>
</tr>
<tr>
<td>Water</td>
<td>0.00 dB/cm/MHz</td>
</tr>
</tbody>
</table>

*Attenuation Coefficients (α) of Common Tissues*
How is the Ultrasound Image Produced?

- Ultrasound images display variations in the acoustic impedance (Z) of different tissues. These are caused by variations in density and compressibility of different tissues, e.g. bone versus liver.

- In its simplest form, an ultrasonic pressure wave is transmitted from the transducer along a single line of sight into the body.

- As the pressure wave propagates, echoes are reflected from objects along the line of sight.

- The returning echoes are received by the transducer.

- Image is displayed by mapping the echo magnitude as brightness in the image, and by mapping the arrival time as distance along the line of sight. This is explained in more detail on the following page.

- This is repeated along many lines of sight to produce a two-dimensional image.

In reality, modern ultrasound machines have different ways of transmitting and receiving multiple ultrasound waves simultaneously, as will be explained later in the module.
Propagation velocity is critical in determining the distance of a reflecting interface from the transducer.

A transducer is any device that converts one form of energy to another. In ultrasound imaging the transducer converts electrical energy into acoustic pulses and transmits it into the body. The transducer then acts as the receiver of the reflected echoes, converting pressure changes into electrical signals which are then processed as an image.

The information that generates an ultrasound scan is based on the precise measurement of time. The depth of the interface that generated the echo can be calculated by measuring the time an ultrasound pulse is delivered until the time the echo returns.

For example, if the time interval from the transmission of the pulse until the return of the echo is 0.04 ms and the velocity of sound is 1540 m/s then:

\[
1540\text{m/sec} \times 0.00005 \text{ sec} = 0.077\text{m} \text{ or } 7.7 \text{ cm}
\]

Since the time measured includes the time for sound to travel to the interface and then return to the transducer, the distance of the interface is:

\[
7.7 \text{ cm} / 2 = 3.85
\]
Instrumentation

Ultrasound Scanners
All ultrasound scanners consist of similar components that perform the same key functions. One of these is a transmitter that sends pulses to the transducer, a receiver and a processor that detects and amplifies the backscattered energy.

The Transmitter
Although the transducer is itself the transmitter of the ultrasound pulses into the body, it must be energized initially by the transmitter, which applies precisely timed, high-amplitude voltage to the transducer. The length of an ultrasound pulse is determined by the number of alternating voltage changes applied to the transducer. Transducers have a range of frequencies which they are able to produce. This is known as the *bandwidth*. 
A Transducer (sometimes referred to as a probe) is any device that converts one form of energy to another.

The ultrasound transducer has a dual function - it is a transmitter that produces ultrasound by converting electrical energy into mechanical energy and then it becomes the receiver, detecting returning ultrasound waves by converting mechanical into electrical energy.

There are many different transducer shapes and configurations depending on which body part is being imaged.

The diagram below shows main components of a simple single-element transducer.

- The crystal is coated with a thin conducting film to ensure good contact with the two electrodes that supply the electric voltage that makes the crystal vibrate.
- Front electrode is earthed to protect the patient from electric shock - has a matching layer to improve the ultrasound transmission.
- Thick backing block at the back face of the crystal absorbs the ultrasound transmitted into the transducer and dampens the vibration of the crystal.
- Plastic housing is insulated by a layer of cork or rubber.
The Piezoelectric Effect

Certain crystals change their physical dimensions when subjected to an electric field, and change back when the field is removed. When compressed, they also have the property of generating electric potentials. Piezo comes from the Greek word, piezein, which means pressure. Changes in polarity of a voltage applied to a transducer cause the transducer to change in thickness, expanding and contracting as the polarity changes. This results in increases and decreases in pressure, producing ultrasound waves that can be transmitted into the body. Pressure changes caused by the returning ultrasound echoes are converted back into electrical energy signals, which are transferred to a computer to create an ultrasound image. **These small events are the source of all ultrasound images.**

Sketch of a crystal that has piezoelectric properties. The crystal changes shape as the surrounding electrical field is reversed. The wavelength ($\lambda$) of the emitted ultrasound is a function of the size of the crystal.
Piezoelectric Materials*

- Transducers are made of a synthetic ceramic (piezoceramic) such as lead-zirconate-titanate (PZT) or polyvinylidene difluoride (PVDF), or a composite.

- Transducer crystals do not conduct electricity, but are coated with a thin layer of silver which acts as an electrode.

- The thickness and acoustic velocity of a piezoelectric crystal determine the resonant frequency of the transducer.

- Transducer crystals usually are made so that their thickness (t) is equal to half the wavelength (\( \lambda \)), so that \( t = \frac{\lambda}{2} \)

- Changing the thickness of the crystal will change the frequency, but not the ultrasound amplitude or velocity, e.g. a crystal with a thickness of 1mm and the velocity of sound is 4000 m/s, then the resonant frequency will be \( f = \frac{v}{\lambda} = \frac{v}{2t} = 2 \text{ MHz} \)

- High frequency transducers are thin, and low-frequency transducers are thicker.

- Transducers can have piezoelectric crystals arranged as either a single source, a multiple crystal source, or a disk source.

The Receiver

- The receiver not only detects, but differentially amplifies weak signals emanating from different depths. Different tissue thicknesses attenuate the ultrasound variably, and the difference in echo strength is compensated by time gain compensation (TGC). (See diagram on following page).

- The attenuation of sound is proportional to the frequency and is constant for certain tissues. Weaker echoes returning from deeper tissues must be amplified more to produce a uniform appearance of the organ in question. The operator can achieve this using the TCG control that allows selective amplification from deeper structures and suppression of signals from more superficial structures. TGC is one of the most important user controls and has an enormous impact on the quality of the image. Some modern ultrasound machines provide automatic TGC.

- Another important function carried out by the receiver is the compression and remapping of the returning data into allowable dynamic range that can be seen on a gray-scale display. The actual range of reflected signals can vary widely. The receiver can selectively amplify returning signals so that they match the dynamic range of the display.
Time Gain Compensation (TGC)

Time Gain Compensation is an amplification technique to increase ultrasound echoes from tissue interfaces that are deeper within the body. This is to compensate for the increasing attenuation of the echoes returning from these deeper areas. This is one of the manual controls available to the sonographer to achieve a more uniform grey-scale image.
The transducer produces ultrasound pulses, making wavefronts that form a three-dimensional beam of ultrasound. The determinants that shape the characteristics of the beam are the curvature of the transducer, the pressure waves, and the acoustic lens that focuses the beam. Changes in the pressure waves produce an area of highly variable amplitude closest to the transducer. This area is called the near field, or Fresnel Zone. Beyond the Fresnel Zone, at a distance that is determined by the radius of the transducer and the frequency, the sound field begins to diverge steadily and the further it gets from the transducer, the more the pressure amplitude decreases. This region is called the far field, or Fraunhofer Zone. This divergence of the ultrasound beam is corrected using a focusing lens, or by using electronic focusing.

Fresnel Zone Depth = \( \frac{R^2}{\lambda} \)

\( R = \) radius of the transducer
All diagnostic transducers are focused, either by using a curved piezoelectric crystal or an acoustic lens. Transducers commonly use an array of piezoelectric elements to transmit a sound pulse into the body and to receive the returning echoes. The transmit signals passing to, and the received signals passing from the array elements can be individually delayed in time, and are then termed a phased array. This arrangement electronically steers and focuses each of a sequence of acoustic pulses through the plane or volume that is imaged, in a process called beamforming. This produces a 2- or 3-D map of the scattered echoes on the display.

The frequency of the ultrasound beam affects the quality of the resulting image to a great degree. Higher frequency ultrasound waves have a longer near field and less divergence in the far field; they permit better resolution of small structures. However, more energy is absorbed and scattered by the soft tissues so that higher frequencies have less penetrating ability. Conversely, a transducer producing lower frequencies will provide greater depth of penetration but less well defined images.
Factors Affecting Beamsteering

a) Changes in Diameter

Diagram demonstrating the influence of transducer size on the ultrasonic beam. The near field is much shorter and divergence considerably greater when the transducer is smaller.
Factors Affecting Beamsteering

b) Changes in Frequency

The effect of transducer frequency on the near field. Higher frequency transducers have much longer near fields.

- 2.25 MHz: Near Field 5.26 cm, Far Field 4
- 3.5 MHz: Near Field 6.2 cm, Far Field 2.6
- 5 MHz: Near Field 11.6 cm, Far Field 1.8
c) Mechanical Focus

Diagrams of the ultrasonic beam emitted by A, an unfocused, and B, a focused transducer.
Factors Affecting Beamsteering

d) Electronic Focusing

Electronic Focused Phase Array

Diagram demonstrating how a phased array transducer can focus the ultrasonic beam. By appropriate timing of the individual elements, the leading edges of the wavelets can produce a concentrically curved wave front so that the resultant ultrasonic beam focuses at a given point from the transducer.
Resolution Performance and Image Quality

Depth (Axial) Resolution

The spatial resolution of an ultrasound device determines its ability to differentiate two adjacent objects as being different and distinct structures. Spatial resolution must be considered in three planes, a) Depth (Axial) Resolution, b) Lateral Resolution, and c) Azimuth or Elevation Resolution, each having different determinants. Elevation resolution is determined by the construction of the transducer and cannot usually be controlled by the user.

a) Depth (Axial) Resolution

Determines the ability to separate two objects lying along the axis of the beam. In pulsed wave ultrasound, the transducer emits a series of brief bursts of sound of about 2 –3 cycles each into the body. The pulse length is the product of the wavelength (which decreases with increasing frequency) and the number of cycles in the pulse. Axial resolution is determined by the pulse length.

Since the pulse length determines axial resolution, higher transducer frequencies give better image resolution.

For example: a transducer operating at 5MHz produces sound with a wavelength of 0.31mm. If each pulse is 3 cycles long then the pulse length is $3 \times 0.31 = 0.93$mm. So 0.93mm is the maximum resolution along the beam axis.

If the transducer frequency is 10MHz, then the wavelength is 0.15mm, producing a resolution of 0.45mm. (See diagram on next page).
Resolution Performance and Image Quality
Depth (Axial) Resolution

Objects A and B in the diagram are 0.5mm apart. A transducer at 5MHz produces a wavelength of 0.31mm and the pulse length (upon which axial resolution depends) is 0.93mm, so A and B cannot be distinguished as separate structures. However, at 10MHz the pulse length is 0.45mm and objects A and B can be differentiated.
b) **Lateral Resolution** is the resolution in the plane perpendicular to the beam and parallel to the transducer. It is determined by the width of the ultrasound beam. Excessive beam widths limit the definition of fine structures such as small cystic areas in atheromatous plaque.

As the beam width varies with depth, so does the lateral resolution.

Lateral resolution is controlled by focusing the beam, either by an acoustic lens, or by a curved piezoelectric crystal. This alters the beam width at selected depths.
Types of Scanner Modes – A-Mode

There are different ways to display ultrasound signals. The three major types are:

- **A-Mode (Amplitude Mode)**
- **B-Mode (Brightness Mode)**
- **M-Mode or T-M Mode (Motion or Time-Motion Mode)**

The earliest displays used A-mode devices. A-mode displays the voltage produced across the transducer on an oscilloscope. The amplitude of the reflected sound is shown by the height of the vertical deflection on the oscilloscope. For example, if a sound wave is transmitted through the side of the head, the reflected beam is a line with three distinct peaks (amplitude) that form where the sound is reflected off three hard formations. One is the skull closest to the transducer, next is the midline structure in the brain, the falx cerebri, which is not as hard as bone, but is hard enough to deflect sound, and the third peak is formed from the skull bone on the opposite side of the head.

As late as the 1960’s, this type of scan (called an echoencephalogram) was used as a crude check for a brain tumour or for bleeding from a vessel within the brain, which could be inferred if the falx was shifted from the midline.
B-Mode refers to Brightness mode. This is the mainstay of ultrasound imaging, providing a real-time, gray-scale display, where the variations in intensity and brightness indicate reflected signals of different amplitudes (the brighter parts indicate larger reflections of sound).

The first B-mode images were simple black or white pictures, with no shades of grey. Grey-scale images were a huge step forward in the quality of ultrasound pictures. In modern ultrasound scanners, the transducer position produces a series of dots of variable brightness on the display screen by sampling multiple lines, which build up a 2-dimensional representation of echoes returning from the different body parts being scanned.

On a black background, the signals of greatest intensity are white, absence of echoes are black, with all the intermediate intensities appearing as shades of gray, with each intensity of the reflected sound being assigned a gray-scale value.

The scanner has a digital memory of 512 x 512 x 512 x 640 pixels, which is used to store the values associated with the echo intensities, which is then sent to the video monitor to display the gamut of shades for that particular ultrasound image. The operator can then adjust the dynamic range of the display, which is best using as wide a dynamic range as possible to differentiate the slightest changes in tissue echogenicity.
I) A short pulse is generated by the ultrasound source (transducer) and propagates into the body at the speed of sound (~1540 m/s).

II) At an interface a small fraction of the pulse (echo) is reflected back to the transducer.

III) The echo signals are displayed as a function of time. This display is called an “A” scan.

IV) When the transducer is coupled to an arm and the signals are displayed as bright spots representing the tomographic image of the tissue, the image is called a “B” scan.
M-Mode (or TM-Mode)

- M-mode [Motion mode - sometimes called TM-mode (Time-Motion-Mode)] provides a one-dimensional view of moving objects over time. It is a particularly useful modality in echocardiography where it displays the echo amplitude from the beating heart, including the motion of the heart valves.

- The transducer is positioned over the heart and is kept stationary. It records the returning echoes over the same line of sight repeatedly. What is changing is the position of the heart wall and valves from one moment to the next. The brightness of the display indicates the intensity of the reflected signal.

- A limitation of M-mode imaging is the difficulty in achieving consistent and accurate beam placement for standard measurements and calculations. Beam placement guided by 2-D imaging can be used but accurate placement of the M-mode beam at the appropriate locations within the heart, as well as on the endocardial surfaces are crucial to obtain accurate measurements and calculations.
Clinical Applications of Ultrasound

The following pages briefly describe how ultrasound is typically used in a clinical setting and then describes in more detail how it used for the vascular system.

ABDOMINAL ULTRASONOGRAPHY
Ultrasonography is often the initial investigation in patients with abdominal pain. Ascites can be identified, and any masses can be evaluated for biopsy. Ultrasonography can also localize areas of sepsis. In acute situations, such as appendicitis, ultrasonography may show an appendix mass, stone, or a focal collection of free fluid next to the appendix. Ultrasound can also diagnose acute cholecystitis and intussusception of the bowel.

OBSTETRICS
Ultrasound is used in early pregnancy to confirm intrauterine and to exclude ectopic pregnancy. A fetus can be detected starting at around seven weeks' gestation and this can be achieved even earlier using transvaginal probes. Later in pregnancy ultrasonography is used to assess fetal growth and to identify any abnormalities. The number of conditions that can be identified prenatally using ultrasonography includes renal abnormalities, diaphragmatic hernias, neural tube defects, and congenital heart defects. There are several ultrasound-guided prenatal interventions which can be performed, such as amniocentesis, chorionic villus biopsy, and intrauterine fetal transfusion.

Ultrasoundography is also excellent at detecting pancreatitis, gall stones and cirrhosis of the liver as well as its complications such as splenomegaly, ascites, varices, and portal hypertension.
URINARY TRACT
Ultrasonography is generally the first choice in conditions such as haematuria, being able to identify tumours in the bladder, renal masses, and bladder outflow obstruction due to an enlarged prostate. Transrectal ultrasonography and biopsy are useful for patients with prostatic malignancy. Prostate volume can also be estimated and correlated with serum concentrations of prostate specific antigen to calculate the likelihood of malignant disease.

CHEST
Ultrasonography is not very useful in the lungs because air causes artifacts to appear in the images. It can be used to locate and drain small effusions in pleural disease and biopsy specimens can be taken under ultrasound guidance.

PAEDIATRICS
Ultrasonography has several specific indications in children. Excellent images of the brain of neonates can be obtained by placing the ultrasound probe on the open fontanelle. Congenital anomalies, including the presence and causes of hydrocephalus can also be shown.

Intussusception and pyloric stenosis can be identified in babies and small children, avoiding the need for contrast media and ionising radiation. Septic arthritis of the hip can be diagnosed from showing effusion into the joint. Ultrasonography is now used to screen infants to exclude congenital hip dislocation at an early and treatable stage.
SMALL BODY PARTS

BREAST
Ultrasonography is not used to screen for breast cancer, as it does not have the sensitivity of mammography. It is used in patients who have a palpable mass which cannot be seen on mammography, because it can differentiate between a fibroadenoma from a cyst (which look identical on mammography). It is also useful in investigating painful lumps such as abscesses, which cannot be compressed for mammography.

SCROTUM
Ultrasonography is the best investigation for scrotal masses and can also be used to identify epididymitis. In testicular torsion ultrasonography may help to show an abnormal lie of the testis and Doppler scanning can be used to check perfusion.

EYE
Ultrasonography of the eye is quick, painless, and simple. The ultrasound probe is placed directly on the closed eyelid, which is first covered in ultrasound jelly to provide good contact. This technique can show retinal detachments and vitreous hemorrhage and is also useful for detecting foreign bodies such as metal splinters in the eye or the retro-orbital tissues.

THYROID
Ultrasonography is used to guide biopsy of thyroid masses. It can differentiate between a multinodular goitre and a homogeneously enlarged gland.
**VASCULAR**

- Abdominal aortic aneurysms can be measured and followed up with conventional ultrasonography.
- Ultrasonography is an effective method of detecting clots and reduced flow in larger vessels of patients with deep vein thrombosis. However, examination of the calf vessels is laborious and time consuming and can be technically difficult, particularly if the leg is swollen.
- Ultrasonography is also used to follow up arterial limb grafts to predict and prevent graft stenosis.
- Carotid artery disease can be evaluated using Doppler ultrasonography.

**ECHOCARDIOGRAPHY**

- Echocardiographic images are displayed on an oscilloscope screen and can be recorded on videotape, paper and/or radiographic film.
- M-mode is very useful in echocardiography, where it is used to evaluate cardiac chamber size, wall thickness, wall motion, valve configuration and motion, and the proximal great vessels. Using ultrasound, anatomic relationships can be determined and information regarding cardiac function can be derived. This technique provides a sensitive method for detecting pericardial and pleural fluid accumulation, and can allow identification of mass lesions within and adjacent to the heart.
Intravascular Ultrasound (IVUS) is used in cardiology as an adjunct to coronary angiography and coronary angioplasty.

Ultrasound imaging displays images of coronary arteries unlike any other imaging technology. The conventional angiogram records images made by injecting arteries with fluid that can be seen using x-ray techniques. Angiography is considered the "Gold Standard" for imaging coronary vessels, but it is very important to view the anatomy within the vessel walls. This allows information on stenosis and plaque deposits within the walls, and can only be achieved using intravascular ultrasound.

In IVUS, the transducer is mounted at the end of a catheter that is introduced into the coronary artery. The catheter is able to:

- Measure the velocity of blood flow within the artery. This helps to determine the extent of a blockage to ascertain the extent to which the heart muscle is compromised by lack of blood flow.
- Show not only the severity of the narrowing produced by a blockage, but also the composition of the underlying atherosclerotic plaque. This information is very important to choose which angioplasty procedure would be optimal in treating the blockage.
- Give the surgeon an immediate assessment of the angioplasty procedure while the catheters are still place. If the procedure must be extended, it can be completed without the patient needing a return visit, with all the attendant stress, anesthesia etc.
Contrast agents are chemical compounds that increase the contrast in medical images. They have been a routine part of clinical X-ray, CT, MRI and radionuclide imaging for several years, but have only recently been developed for use in ultrasound imaging.

Most ultrasound contrast agents form tiny microbubbles, which are smaller than red blood cells, which allows them to pass through the smallest capillaries of the body. The compound is injected intravenously and stays within the vascular system and is cleared from the body by exhalation from the lungs in a matter of minutes from the initial injection. The microbubbles vibrate in a nonlinear fashion, producing harmonic signals that can be detected by ultrasound systems.

The major application of contrast agents for ultrasound has been in echocardiography, which relies on excellent visualization. Echocardiographic contrast agents image blood flow in the heart in different modes, including colour and power Doppler, and ultrasound angiography. These imaging modalities are explained in the next module.

Research in ultrasound contrast agents for other applications is ongoing and has also spurred the development of enhanced ultrasound equipment, necessary for delineating the improved definition available through contrast agents.

As microbubbles can be destroyed by the ultrasound beam, researchers are investigating the possibility of delivering drugs to targeted tissues by filling microbubbles with drugs, injecting them directly into the target site and then destroying the microbubbles, which will release the drug.