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The physics of ultrasound and Doppler

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Introduction

Ultrasound imaging is an important diagnostic and intervention tool that incorporates high-frequency sound waves to generate images. The Doppler effect, and the different Doppler modalities, form an essential component of ultrasound, enabling the measurement of blood flow velocity and direction. In this article, we will delve into the physics behind ultrasound and Doppler and explore the underlying principles that make these technologies possible.[1]

Sound waves

Waves carry energy from one location to another and may present in the form of sound, heat, magnetic, or light energy. As sound travels through a medium, it results in the vibration of molecules, creating high-pressure compressions and low-pressure rarefactions.[1] Sound travels through a medium in a straight line, which can be represented as a sine wave, as shown in Figure 1.[2]

The definition of ultrasound is the vibration of sound at a frequency above the threshold of human hearing. Table I depicts the speed of sound travelling through different media. Medical ultrasound imaging typically uses sound waves at frequencies of 1–20 megahertz (MHz). Sound wave properties are described in terms of frequency/Hertz (Hz), wavelength (mm), amplitude (dB), power (Watt), intensity (Watt/square cm), period (ms), and propagation velocity (m/s). These properties are depicted in Figure 1 and can be defined as follows:

- Frequency refers to the number of repetitions or sound wave cycles per second and the speed of the sound wave/wavelength (frequency and wavelength are inversely proportional).[1] The speed of sound through soft tissue is 1 540 m/s.[3]
- Wavelength is the distance between excitations or the length of a single wave cycle.
- Amplitude or loudness of the excitation is the energy of the sound wave.[1]

Sound waves, and the media through which they travel, undergo different interactions. The resistance to ultrasound propagation is referred to as acoustic impedance and depends on the medium's density and the speed at which sound travels through

Table I: Speed of sound through different media[1]

Tissue type	Speed (m/s)
Air	330
Lung	500
Fat	1 450
Liver	1 560
Blood	1 560
Muscle	1 600
Tendon	1 700
Bone	3 500
Soft tissue (average)	1 540

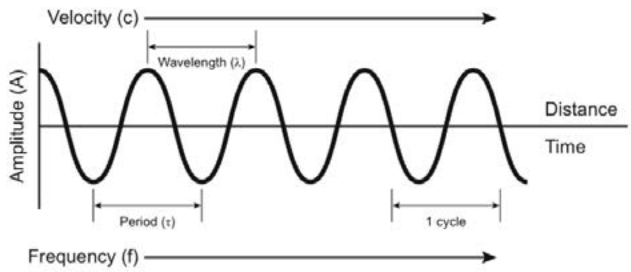

Figure 1: Different aspects of a wave[1]

that particular medium.[1] Ultrasound waves encounter interfaces between two different tissue media and may undergo reflection, transmission, absorption, attenuation, scattering, refraction, and diffraction.[1,3] These terms can be defined as follows:

- Reflection occurs when waves bounce back at the interface. The greater the difference in acoustic impedance across a tissue boundary, the more ultrasound will be reflected.
- Transmission happens when the waves pass through the interface.
- Absorption occurs with the conversion of ultrasound energy to heat.
- Attenuation refers to the reduction in intensity and amplitude of the ultrasound waves as they move through a medium and energy is lost. Attenuation is caused by absorption, scattering, and divergence. Different tissues and media have different attenuation coefficients, enabling the detection of structural

differences in ultrasound imaging. Echoes are generated when ultrasound waves encounter tissue interfaces. These echoes are received by the transducer and converted into electrical signals and images.[1]

- Scattering occurs when ultrasound beams strike an object at right angles and are scattered in all directions. If the tissue has an irregular border, the beams will not be returned to the transducer and most of the energy is lost to surrounding tissues.
- Refraction refers to a sound wave's change in velocity through media with different acoustic impedance, resulting in a change in the direction of the ultrasound wave.
- Diffraction refers to the spreading out of a beam as it moves farther from the ultrasound source and as the intensity of the ultrasound beam decreases.

Ultrasound probes and the piezoelectric effect

Ultrasound transducers consist of piezoelectric crystals (e.g. lead zirconate titanate) that convert electrical energy into sound waves and sound waves back into electrical energy. When an alternating current is applied to the transducer, the ultrasound crystals deform and produce ultrasound waves. The piezoelectric effect is the vibration of a piezoelectric crystal at the tip of the transducer that generates a specific ultrasonic frequency to create ultrasound waves. The ultrasound waves penetrate the propagation medium and are reflected to return to the transducer. The transducer then converts the returning ultrasound waves to electrical energy, which can be interpreted as an image on the ultrasound machine.[1,2]

Image quality is determined by tissue resolution (the ability to distinguish two objects that are spatially close together) and penetration. High-frequency probes have decreased penetration because the piezoelectric crystal can only send out a limited number of ultrasound waves before the waves dissipate.[1] Probes that emit higher frequencies and lower wavelengths result in better resolution but compromised penetration.[3] Table II depicts the properties of the different ultrasound probes.

Table II: Properties of the different ultrasound probes[3]

Transducer type	Phased array	Linear	Curvilinear
Frequency	1–5 MHz	5–10 MHz	2–5 MHz
Depth	35 cm	9 cm	30 cm
Applications	Heart, lungs	Abdominal, pelvic	Arteries, veins, musculoskeletal, eyes, and testes

Imaging modes

Ultrasound systems offer different imaging modes, each with unique capabilities and applications.

A-mode (amplitude mode)

This displays a single echo signal against time to measure depth.[1] It is rarely used in today's clinical practice.

B-mode (brightness mode)

B-mode ultrasound is the most used mode in diagnostic ultrasound. B-mode imaging uses the amplitude of the received ultrasound echoes to represent different shades of grey on the image. This allows for the visualisation of different tissues and structures. B-mode provides two-dimensional (2D) images and has a wide range of applications.[2]

M-mode (motion mode)

M-mode ultrasound is a specialised mode that displays a one-dimensional (1D) representation of motion over time. It is primarily used to assess dynamic structures, such as cardiac motion. In M-mode imaging, a single ultrasound beam is continuously transmitted at a fixed location. This generates a real-time image of motion along a selected scan line.[2]

Three-dimensional (3D) and four-dimensional (4D) ultrasound modes

3D and 4D ultrasound modes provide a more comprehensive view of anatomical structures. 3D imaging captures multiple 2D slices of a region, which are then reconstructed to form a 3D volume. 4D imaging adds real-time motion to 3D imaging, which allows for the visualisation of dynamic parameters and structures.[1]

Doppler mode

Doppler ultrasound is an ultrasound mode that utilises the Doppler effect to measure the velocity and direction of blood flow. It provides information about blood flow patterns, and valvular pathology, enabling the assessment of haemodynamic parameters. Doppler ultrasound can be further classified into colour and spectral Doppler.[1]

Colour Doppler imaging incorporates the overlaying of colour-coded representations of flow information onto the B-mode image. This allows for the visualisation of blood flow direction and velocity. Different colours (e.g. red and blue) represent flow towards and away from the transducer.[1,2]

Spectral Doppler provides quantitative information about blood flow velocities. It displays the Doppler frequency spectrum as a graph, known as the Doppler spectrum or waveform. Spectral Doppler is used in the evaluation of cardiac function.[2]

The Doppler effect

The Doppler principle is a fundamental concept in physics also applied in medical imaging, such as ultrasound. In ultrasound, the Doppler effect enables the measurement of blood flow velocity and direction. The Doppler principle was named after Christian Doppler. It describes the change in the frequency of a sound wave when the source or observer is in relative motion to one another.[2,3] This shift in the frequency of the wave is known as the Doppler frequency shift. The frequency shift is directly proportional to the velocity.

In medical practice, the frequency shift occurs when ultrasound waves encounter moving blood cells and the frequency of the reflected waves changes. The ultrasound will pick up this shift in frequency. If an object moves toward the transducer, the frequency increases (positive Doppler shift), while movement away from the transducer results in a decreased frequency (negative Doppler shift).[1,2] The Doppler principle can further be explained using the wave equation. When a wave source and an observer are in motion relative to each other, the observed frequency differs from the emitted frequency according to the equation set out below.[1]

Doppler equation

The Doppler equation relates the observed frequency shift to the velocity of the moving blood cells, the speed of sound in tissue, and the angle between the ultrasound beam and the direction of blood flow. The equation is given by:

$\Delta f = 2 \times f \times V \times \cos a / c$

Where:

Δf = Doppler frequency shift (Hz)

f = ultrasound frequency (Hz)

V = velocity of blood flow (m/s)

d = direction of blood flow relative to the ultrasound beam = cos a (Cosine of the angle between blood flow direction and ultrasound beam.)

c = speed of sound in tissue (1 540 m/s)

The sign convention depends on whether the observer and source are moving toward each other or away from each other.

Continuous wave (CW) Doppler

CW Doppler utilises two separate ultrasound transducers. One transducer is used for transmitting and the other for receiving. This allows for continuous emitting and detecting of sound waves. The frequency shift is measured along the entire ultrasound beam. This provides information about blood flow velocity along the entire beam and cannot identify its location. CW Doppler is commonly used in cardiovascular applications requiring measurement of high velocities.[1,2]

Pulsed wave (PW) Doppler

PW Doppler employs a single transducer that alternates between transmitting and receiving ultrasound waves. It allows the user to select a specific region of interest and enables depth-specific measurements, thus providing information about blood flow velocity at different locations. PW Doppler is widely used in clinical applications, including obstetrics, cardiology, and vascular imaging, and can measure lower velocities than CW Doppler.[2,3]

Nyquist limit and aliasing

The Nyquist limit is a fundamental concept in Doppler ultrasound that allows for the accurate measurement of blood flow velocity. Understanding the Nyquist limit is essential for ensuring accurate and reliable measurements.[1]

The Nyquist sampling theorem is a fundamental concept in signal processing. To accurately reconstruct a continuous signal, it states that the signal must be sampled at a rate at least twice the highest frequency component present in the signal.[1] This principle applies to Doppler ultrasound as well. According to the Nyquist theorem, the maximum measurable velocity is limited by the ultrasound system's pulse repetition frequency (PRF). The Nyquist limit states that the PRF must be at least twice the maximum Doppler frequency shift to avoid aliasing. The Nyquist limit is the upper limit of detectable velocities without aliasing. Aliasing occurs when the Doppler frequency shift exceeds the Nyquist frequency, leading to an incorrect representation of blood flow velocity.[1] Aliasing is where spectral overlap occurs and manifests as a wrap-around effect. This is where high velocities appear as lower velocities in the opposite direction.[1,2]

Ultrasound system configuration

To avoid aliasing and accurately measure blood flow velocities, the ultrasound system must be appropriately configured. The following considerations are important:

a. Doppler angle correction: The Doppler angle is the angle between the ultrasound beam and the direction of blood flow, and this will affect the measured velocity. As the angle increases, the velocity calculations become more inaccurate.[1]
b. Adjusting scale and baseline: The scale and baseline settings determine the range and zero reference for Doppler velocity measurements. Proper adjustment ensures that the desired velocities are within the detectable range and displayed optimally.[1]
c. Doppler gain optimisation: Doppler gain settings affect the sensitivity of the Doppler system. The gain optimisation allows for the clear visualisation of low-velocity flows and avoids excessive artefacts.[1]
d. Colour Doppler mapping: Colour Doppler provides a visual representation of blood flow direction and velocity. The Nyquist limit should be set appropriately to prevent aliasing. The colour scale should also be adjusted to display the desired range of velocities to be measured.[1]

Conclusion

Ultrasound is sound with a frequency above 20 kHz. The physical properties of sound waves (wavelength, frequency, and velocity) dictate the limits of clinical ultrasound. Ultrasound is generated using piezoelectric materials. The image created is dependent on the physical properties of the tissue being examined as well as the ultrasound wave and probe properties.[3] Ultrasound is an extremely complex and useful tool and having a good understanding of all the properties and functions thereof will allow one to utilise the platform as accurately as possible.

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