The physics of ultrasound

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Abstract

Ultrasound is a safe, non-invasive imaging modality which is increasingly used in anaesthesia to aid placement of central venous cannulae and local anaesthetic blocks. Transoesophageal echocardiography (TOE) is used to assess myocardial and valvular function during anaesthesia or on the intensive care unit. It is essential to understand the underlying principles of how the ultrasound image is created in order to optimize the image, and recognize and prevent artefacts.

This article describes the physics of waves and their interactions, and applies these principles to explain how the ultrasound machine produces an image. The Doppler effect and its application to measurement of blood flow and cardiac output is described.

Keywords Artefacts; Doppler effect; piezoelectric effect; imaging; ultrasound; waves

Learning objectives

After reading this article, you should be able to:

- describe the general properties of waves, and how they interact with each other and their environment
- explain how the ultrasound machine produces an image, and how artefacts are formed and prevented
- explain the Doppler effect, and how it may be used clinically to measure blood flow

Properties of waves

A travelling mechanical wave is a disturbance with a regularly repeating and progressively moving profile which propagates energy through a medium. Energy is transferred with no net displacement of the medium. In a **transverse wave** (e.g. ripples on water, electromagnetic radiation), the disturbance is perpendicular to the direction of travel of the energy. In a **longitudinal wave** (e.g. a row of cars colliding, or sound waves in liquid or gas), the disturbance occurs in the same direction as the direction of travel of energy, resulting in oscillating regions of compression (increased density) and rarefaction (decreased density).

From a graph of displacement against distance (Figure 1a), we can define wavelength (λ) as the distance between corresponding points on successive waves, amplitude (a) as the maximum displacement from the equilibrium position, and velocity (c) as how rapidly the energy moves through the medium. From a graph of displacement against time (Figure 1b), the distance between corresponding points on successive waves defines the period (T). The reciprocal of the period is the frequency (f) (i.e. $\boldsymbol{f = 1/T}$) and has derived SI units Hertz (Hz; per second). Alternatively, the frequency of a travelling wave is the number of peaks which pass a stationary observer per second. Velocity, frequency and wavelength are related by the formula: $\boldsymbol{c = f}\lambda$.

Waves of identical frequency may start their oscillations simultaneously (**in phase**), or at different times (**out of phase**); described by the **phase angle** (θ) between their vectors.

Electromagnetic waves can travel across a vacuum, but sound waves require a medium for propagation. The speed of sound (c) is fixed for any given medium. Dense, rigid materials (e.g. bone: $c \approx 3000$ m/s) transmit sound faster than light, compressible materials (e.g. air: $c \approx 330$ m/s). The speed of sound in the soft tissues of the body ranges from ~1400 m/s (fat) to ~1750 m/s (tendon), with an average speed of 1540 m/s.

Human hearing can detect sound in the range 20 Hz to 20 kHz (the upper limit decreases with age; presbyacusis). Frequencies below and above this range are called infrasound and ultrasound respectively. The amplitude of a sound wave determines the flow of energy or intensity (W/m^2), and is frequently expressed on a logarithmic scale as effective sound pressure relative to the threshold of hearing at 1 kHz in non-SI units decibels (dB).

Interactions of waves

If two waves of identical frequency, amplitude and phase combine, a single wave with twice the amplitude results (**constructive interference**). If the waves are 180° out of phase they will cancel each other out (**destructive interference**). Waves interact with their environment. They can bounce (**reflection**), bend (**refraction**), scatter (**diffraction**), and convert their kinetic energy into heat energy (**absorption**). Sound intensity decreases exponentially with distance from the source (**attenuation**; dB/cm) due to diffraction and absorption. Greater attenuation occurs at high frequencies and in light compressible media.

The **acoustic impedance** (z) (derived SI units: Rayl; kg/m^2/s) of a medium is analogous to electrical impedance, and depends on its density and stiffness. When a wave meets an interface between two media of different z, some of the wave's energy will be reflected, and the remainder will be transmitted. The proportion of wave energy reflected depends on the difference in z, and is described by the **amplitude reflection co-efficient**, R_A. For interfaces between most body tissues, around 1% of the beam is reflected (R_A ~0.01); however at an air–tissue interface >99% of the incident beam is reflected (R_A ~0.999) which results in the ultrasound machine being 'dazzled' by the reflected beam and unable to 'see' deeper structures. A **coupling medium** (mineral oil based gel) is therefore necessary to eliminate air between the ultrasound probe and the skin surface.

The ultrasound machine

In its simplest form, the ultrasound machine operates like a ship's depth-sounding device. Periodic electrical pulses are converted into ultrasound pulses by a **lead zirconium titanate (PZT) piezoelectric transducer**. The sound waves travel through the tissues until they meet a tissue interface, whereupon a proportion of the emitted signal (determined by R_A) is reflected back to the transducer, which converts the reflected waves back into electrical pulses which are amplified and displayed. The

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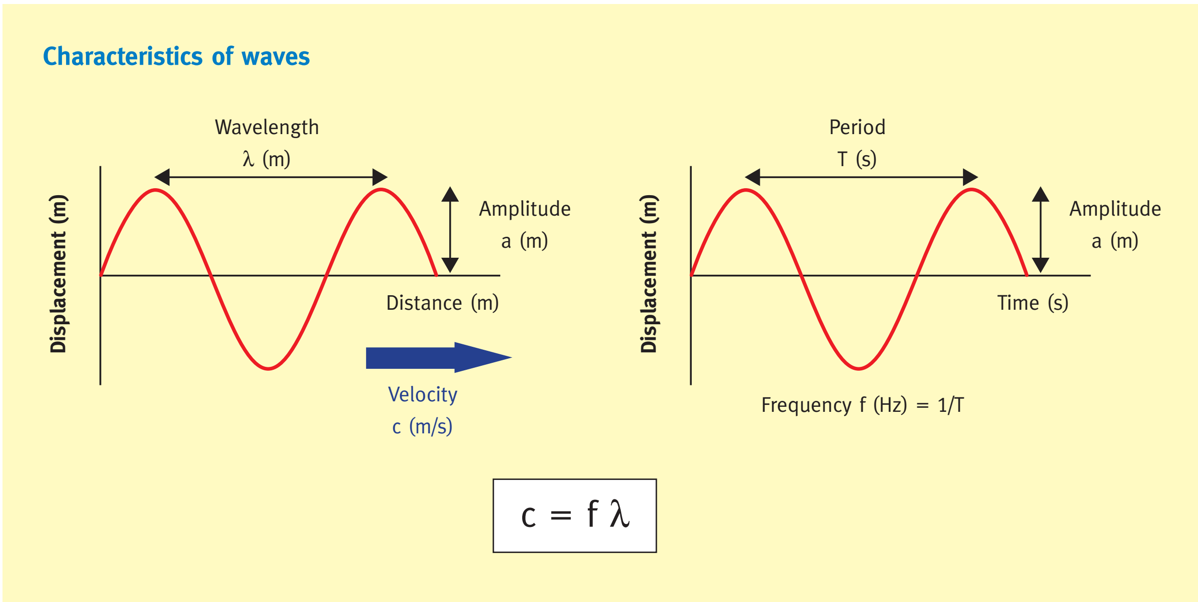

Figure 1

amplitude of the reflected wave provides information about the nature of the tissue interface. If the time taken for the sound to travel to the target and return to the transducer (t) and the speed of sound through the tissues ($c \sim 1540$ m/s) are known, then the depth of the target (d) is given by:

$$d = ct/2$$

An ultrasound probe typically emits sound waves in bursts or pulses of 1 μs duration spaced 1 ms apart, with a **pulse repetition rate** of 1000 pulses per second; so it transmits 1% of the time and 'listens' for the reflected echoes 99% of the time (Figure 2).

The earliest ultrasound machines simply plotted the amplitude of the reflected wave against time on an oscilloscope ('Amplitude'

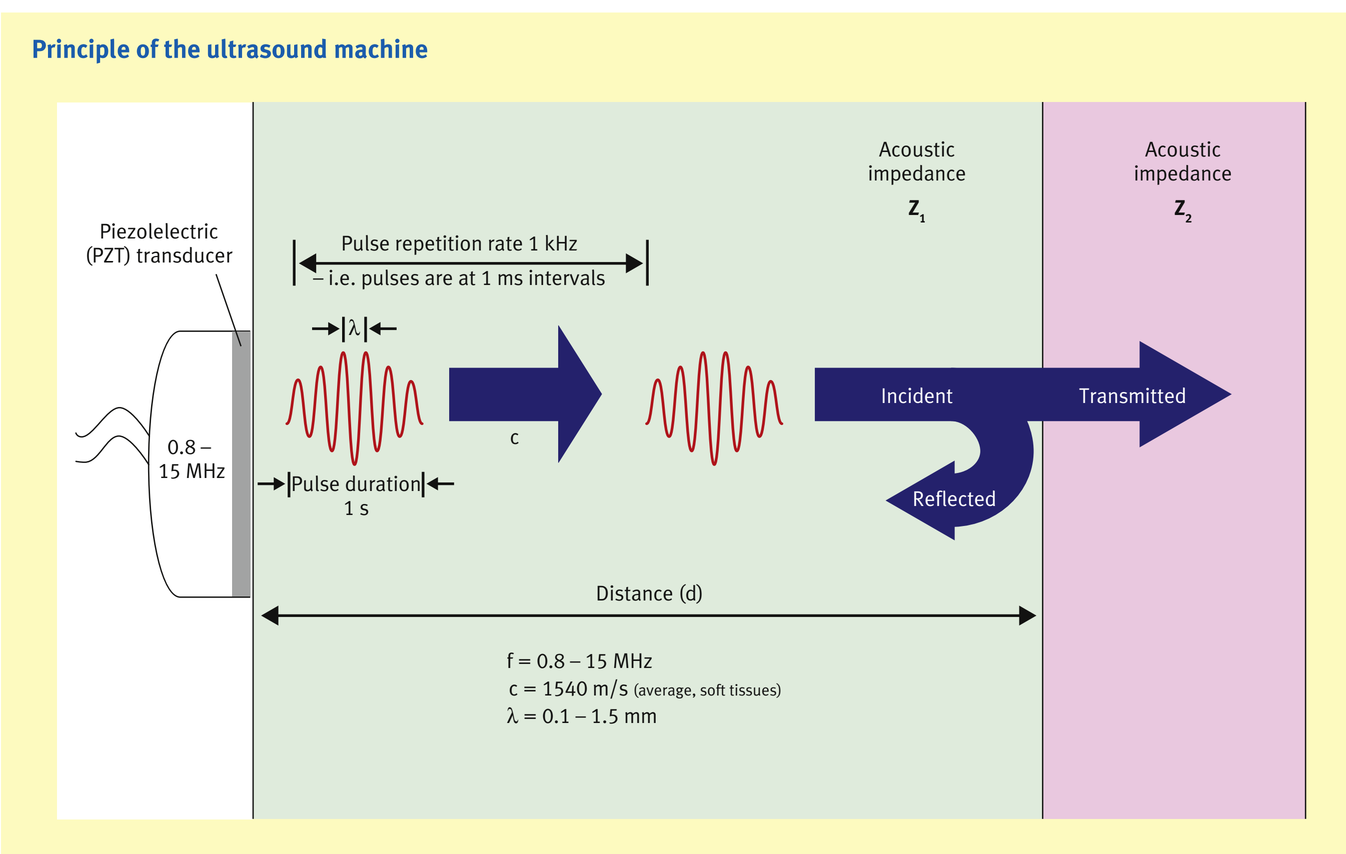

Figure 2

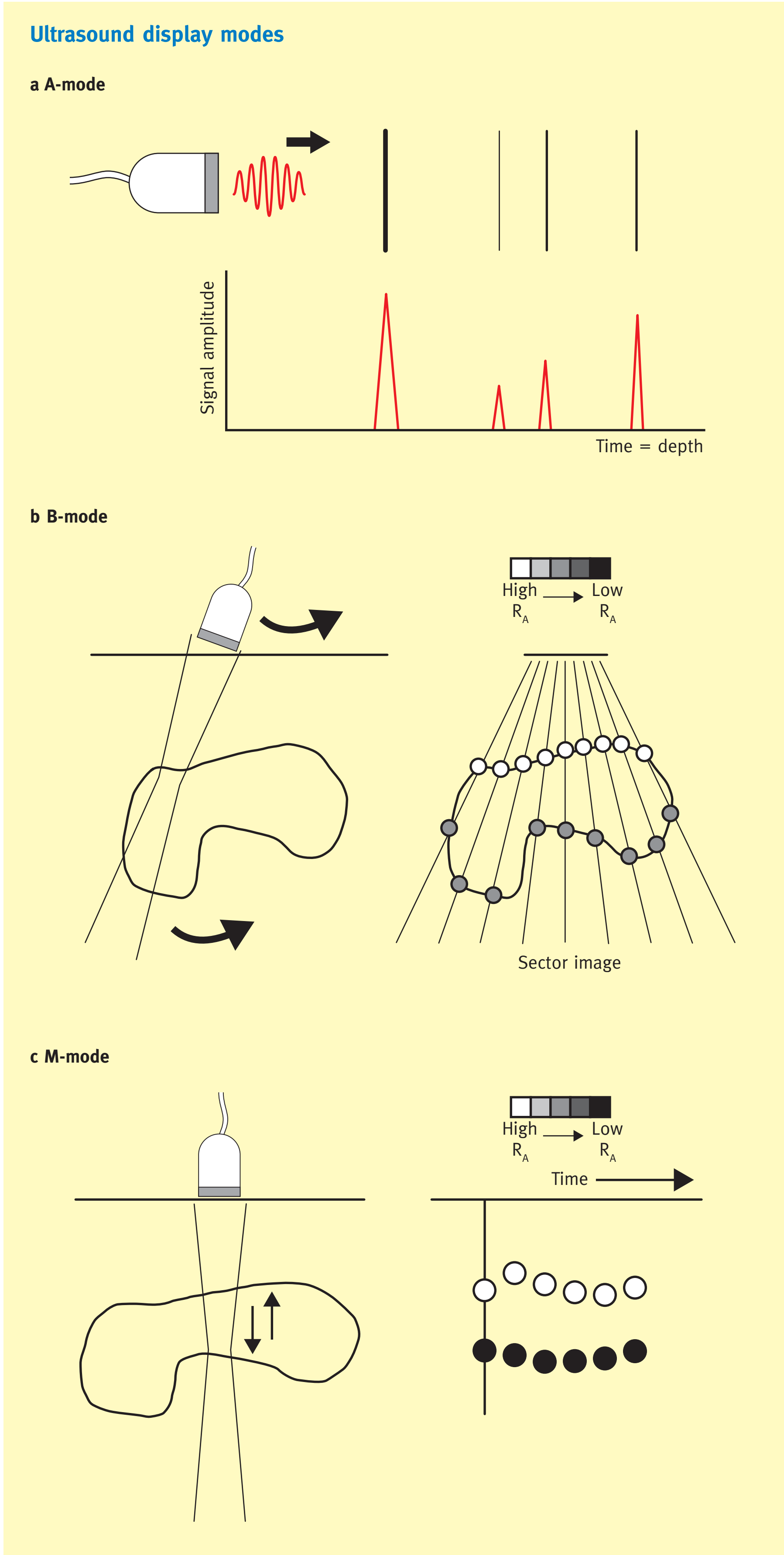
Ultrasound display modes
a A-mode
Signal amplitude
Time = depth
b B-mode
High
R_A
Low
R_A
Sector image
c M-mode
High
R_A
Low
R_A
Time

Figure 3

or **A-mode**). The distance along the time axis to each voltage spike was proportional to the distance from the probe to the interface, and the height of each spike was proportional to R_A at each interface. Alternatively, the amplitude of the reflected signal can be represented as the brightness of pixels against time (= distance) on a computer screen ('brightness modulation' or **B-mode**). A two dimensional (2D) image (**linear** or **sector scan**) can then be formed by translating or rotating the probe mechanically, or electronically with a **phased array** of sequentially activated transducers. If the probe is stationary but the underlying tissue is moving (e.g. a heart valve), a plot of pixel brightness against time will show movement of tissue boundaries with time ('motion' or **M-mode**) (Figure 3).

Getting the best image

It is important to optimize gain, depth and resolution, and avoid and recognize artefacts.

The amplitude of the emitted sound waves decays exponentially en route to the target due to attenuation. Typically only 1% of this signal is reflected at the tissue interface, and this then undergoes further exponential attenuation as it travels back to the transducer. In order to detect the reflected signal, the ultrasound machine must incorporate very powerful amplification with **time gain compensation** (**TGC**), that is, gain increases exponentially from the time that the pulse is emitted, resulting in a roughly constant output signal, regardless of target depth. Incorrect TGC settings may result in horizontal banding artefacts. The field of view must be optimized for the target. If the depth setting is too shallow, the target will not be visible; if it is too deep, the target will appear too small.

A structure which is smaller than the wavelength that is being used to observe it cannot be visualized. Visualization or differentiation between adjacent objects in the direction of the ultrasound beam (**axial resolution**) requires smaller wavelengths, and hence higher frequencies. As $c = f\lambda$ and $c \approx 1540$ m/s, a probe frequency of 1–15 MHz is required to visualize objects of 1 mm diameter. High frequencies are rapidly attenuated, so there is a trade-off between axial resolution and depth penetration. It is therefore important to select a probe of appropriate frequency for the size and depth of the target.

It may be difficult to differentiate between two objects which are side by side in the ultrasound beam (**lateral resolution**) because the reflected waves from each may spread due to diffraction and may undergo interference to form a single wave which appears to emanate from a single object. The ultrasound beam is approximately parallel for only a short distance in the near field (**Fresnel zone**); beyond which diffraction causes the beam to diverge (**Fraunhofer zone**), causing poor lateral resolution. The extent of the Fresnel zone and angle of spread of the Fraunhofer zone are determined by the width and wavelength of the ultrasound beam; but lateral resolution can be improved by focusing the beam with a curved transducer, acoustic lens, or phased array of transducers (**Huygen's principle**).

If the 2D image is refreshed at more than 20 Hz, **persistence of vision** facilitates display of real-time moving images. However, increasing sector size, lines per sector (**line density**) and frames per second (**frame rate; temporal resolution**) all require more pulses to be transmitted, received and processed every second, requiring massive processing power, and may result in poor quality or jerky images (**temporal artefacts**).

The depth of the target is calculated based on the assumption that sound travels through soft tissues at an average speed of 1540 m/s; however if the beam travels through a large amount of tissue with a different velocity of propagation (e.g. fat $c \sim 1420$ m/s), the calculation of depth will be incorrect and the target will be displayed as more deep or superficial than it actually is (**ranging artefact**).

A highly reflective surface (e.g. a calcified atherosclerotic plaque), may prevent the beam from penetrating further, and deeper structures will be in an **acoustic shadow**. Conversely, a fluid-filled cyst may act as an acoustic lens and focus the beam on structures which lie immediately deep to it (**post-cystic enhancement**). Generally reflection is **specular** ('mirror -like'); however if the target size is similar to the wavelength of the beam (e.g. the complex cellular architecture of the liver), **diffuse reflection** and scattering occurs, causing a characteristic speckled appearance. If the beam encounters a highly reflective large smooth surface (e.g. diaphragm) and is not perpendicular to the surface, a **mirror artefact** can occur where the target appears to be in a completely position. A number of parallel highly reflective boundaries (e.g. bladder) may cause multiple overlapping reflections (**reverberation artefact**). If the beam does not strike a tissue interface perpendicular to the boundary, it will bend according to **Snell's law of refraction**, causing the target to appear in the wrong place (**refraction artefact**), for the same reason that a fisherman trying to spear a fish may miss his target due to the refraction of light by the air–water interface.

Doppler ultrasound

Background

The Doppler effect (C. J. Doppler, 1842) is the apparent change in frequency of a wave due to relative motion (velocities) of the wave source (v_s) and observer (v_o); or in the case of a reflected wave, motion of the reflector. The **Doppler shift frequency f_D** (often abbreviated to the 'Doppler shift') is the difference between the actual (transmitted) frequency (f) and observed (received) frequency (f').

$$f' = f\left(\frac{c + v_o}{c - v_s}\right) \qquad\qquad f_D = f - f'$$

Red blood cells provide moving targets for reflection of an ultrasound beam. If a source frequency of 10 MHz is used, and blood flow is 10–30 cm/s, $f_D \approx 200$ Hz, which is within the audible range. This can be clearly heard as a 'whooshing' sound from the loudspeaker of a Doppler flow meter, providing a qualitative indication of perfusion (e.g. patency of peripheral vessels or grafts). 'Duplex' scanners (e.g. TOE and CardioQ®) combine Doppler flow measurement with 2D pulse-echo imaging of vessel cross-sectional area to quantify aortic blood flow and estimate cardiac output.

Display and artefacts

Doppler flow measurement is a cosine function of the **insonation angle** (θ). The probe should ideally be directed along the axis of flow ($\theta = 0°$); impossible in practice, since the probe would have to be inside the vessel. As θ increases, measurements rapidly become increasingly inaccurate, until the probe is perpendicular to flow, when no measurement can be made. For practical purposes, θ should be under 60°.

Doppler flow information is usually displayed on the ultrasound machine in two ways:

Spectral display shows a histogram of the reflected signal against f_D, and provides detailed information about the range of flow rates at one site with time (good **temporal resolution**).

Colour flow display gives poorer temporal resolution and less accurate information about flow, but covers a wider area, and may be superimposed on a B-mode image to give an indication of flow rate and direction. Interpretation of the signal is aided by arbitrary user-definable **colour mapping** (e.g. red and blue may indicate flow towards and away from the probe respectively, or vice versa; and the brightness of each pixel corresponds to the flow velocity at each point).

Low sampling frequencies can result in apparent reversal of the direction of flow. The effect (**Nykvist phenomenon**) is similar to that observed when running water appears to flow backwards when viewed with a stroboscopic light, or when wagon wheels in old movies appear to revolve slowly backwards. This artefact appears on spectral displays as a 'wrapping round' of histogram peaks, and on colour flow displays as patches of red and blue within the same vessel; and is addressed by increasing the pulse repetition frequency. ◆

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