7

Ultrasound Imaging

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INTRODUCTION

Sound is the physical vibration of particles in a medium — *not* electromagnetic (EM) radiation (e.g., light and X-rays). Like light (but unlike X-rays and gamma rays), sound can be reflected, refracted, and focussed. Unlike electromagnetic waves, sound requires a medium in which to travel.

Ultrasound is sound with a frequency above the human audible range (>20 kHz). Audible sound spreads through a room, whereas the short wavelengths of ultrasound allow it to be directed into a beam. Advantages of ultrasound over X-ray imaging include superior soft tissue contrast, and safety — for example in obstetric imaging — due to perceived comparatively insignificant hazards.

Longitudinal compression waves travel in the direction of particle movement (see Fig. 7.1). Particles moving closer together result in increased pressure (compression) and particles moving apart result in reduced pressure (rarefaction). Normal pressure at 1 atm is 101 kPa. Longitudinal waves are those used in standard ultrasound imaging. Transverse shear waves travel in the direction perpendicular to particle motion (Fig. 7.2). This type of wave can be used in shear wave elastography.

PHYSICAL PROPERTIES OF ULTRASOUND

The **frequency (*f*)** of a sound wave is how many times per second the compression phase passes any single point in the medium, measured in megahertz (MHz). The **period (*T*)** of a wave is the *time* between successive compressions (or rarefactions) *at a single position in the medium* (Fig. 7.1A).

$$T = \frac{1}{f}$$

The **wavelength (λ)** of a wave is the *distance* between successive compressions (or rarefactions) *at a single point in time* (Fig. 7.1B). The **speed of sound *(c)*** depends on the medium through which it travels. In tissue, ultrasound behaves as in a fluid, so the speed of sound

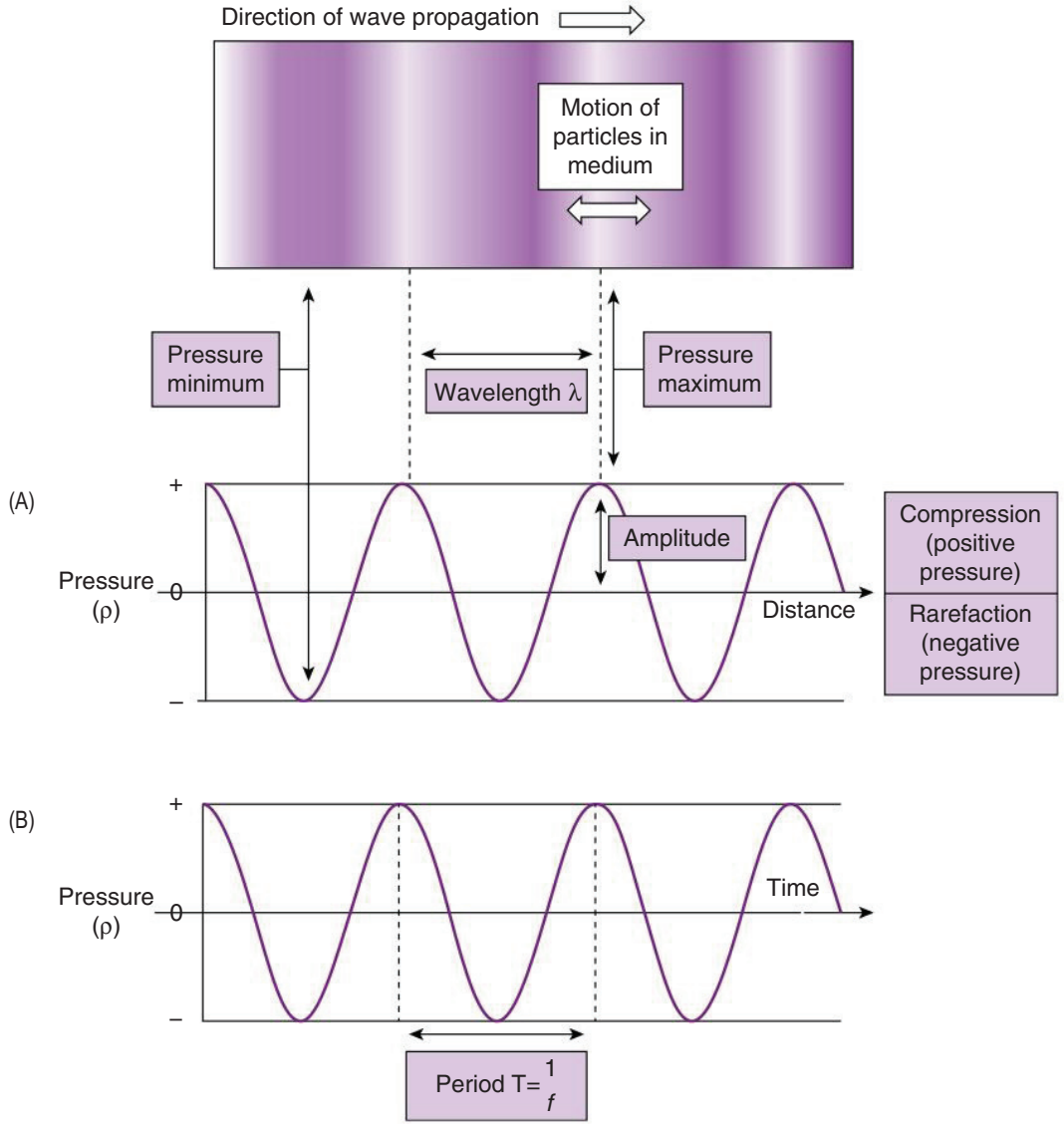

Fig. 7.1 Propagation of a longitudinal pressure wave and graphs of the sinusoidal continuous waves showing excess pressure (A) versus distance and (B) versus time.

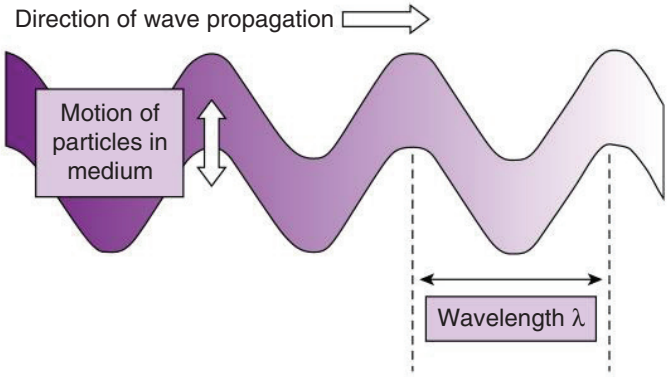

Fig. 7.2 Propagation of a transverse shear wave.

depends on the stiffness (adiabatic elastic bulk modulus, B) and density (ρ):

$$c = \sqrt{\frac{B}{\rho}}$$

Hence, sound speed increases with the stiffness and decreases with the density of the medium. It also increases with temperature. Table 7.1 shows the density and speed of sound in various media. Air has a much lower density, but is much more compressible than water or tissue, hence the low velocity.

For practical purposes, sound travels through a homogeneous medium at a constant speed c that is independent of frequency and wavelength:

$$c = f\lambda$$

TABLE 7.1 Properties of Ultrasound in Various Media

Material	Speed *(c)* (m/s)	Density *(ρ)* (kg/m^3)
Air	330	1.29
Average soft tissue	1540[a]	1000
Typical bone	3200	1650
Lead zirconate titanate (PZT)	4000	7500

[a]Range of speed: 1300–1800 m/s

TABLE 7.2 Acoustic Impedances of Various Media

Tissue	Acoustic Impedance *(Z)* (kg/m^{2}s)
Air	430
Muscle	1.70×10^6
Liver	1.64×10^6
Spleen	1.63×10^6
Kidney	1.62×10^6
Average soft tissue	1.5×10^6
Fat	1.38×10^6
Typical bone	5.3×10^6
Lead zirconate titanate (PZT)	30×10^6

The intensity of ultrasound, measured in watts per square millimetre (W/mm^2), is proportional to the square of the wave amplitude (Fig. 7.1A) and is under the operator's control.

Interference is the interaction that occurs when two waves cross each other (Fig. 7.3).

- Constructive interference occurs when the two waves are exactly in step (in phase) and their amplitudes add up.
- Destructive interference occurs when the two waves are out of phase. This results in a reduction in intensity. If they are equal and exactly out of phase, they completely cancel out.

Acoustic impedance (*Z*) describes the resistance experienced by an ultrasound beam in the medium. It depends on density (ρ) and elasticity and is, for practical purposes, independent of frequency. Acoustic impedance is measured in Rayls (kg/m^{2}s) and can be calculated using the speed of sound in the medium:

$$Z = \rho c$$

Examples of impedance in various media are given in Table 7.2. The fraction of sound energy reflected and transmitted between two media depends on the acoustic impedance.

PIEZOELECTRIC EFFECT

Piezoelectric materials convert electrical energy into sound energy and vice versa (Fig. 7.4). These allow charge accumulation after the application of mechanical stresses. Such materials used in ultrasound include lead zirconate titanate (PZT) composite or plastic polyvinylidine difluoride (PVDF). When heated above the Curie temperature (individual for each material; e.g., 350°C for PZT), the piezoelectric properties are lost.

In an ultrasound transducer, two opposite faces are coated with electrically conducting silver. To produce an

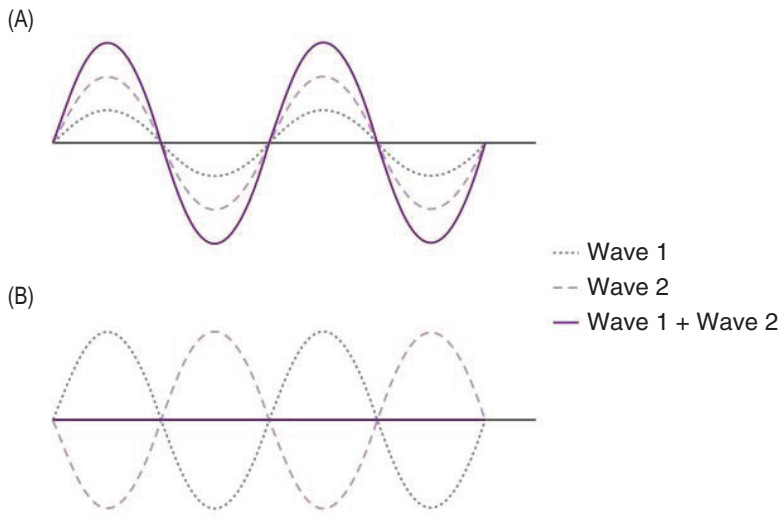

Fig. 7.3 Ultrasound interference: (A) constructive and (B) destructive.

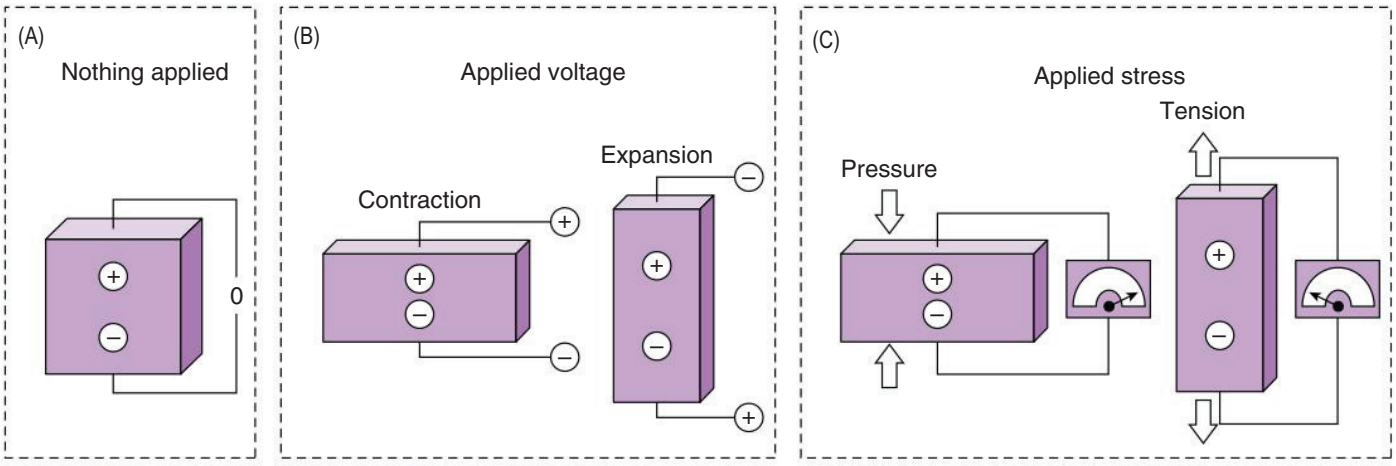

Fig. 7.4 Piezoelectric material with (A) nothing applied; (B) a voltage applied across the transducer, causing a mechanical oscillation; and (C) a stress applied, producing a voltage across the transducer.

ultrasound wave, a voltage is applied across the piezoelectric material, causing it to expand or contract (Fig. 7.4B). This effect is proportional to the voltage and reverses if the voltage is reversed. When coupled with a medium, this creates a pressure wave, which travels through the medium.

To detect ultrasound: the incoming pressure wave creates a voltage across the transducer (Fig. 7.4C), which can be detected. Hence, the transducer converts sound energy to electrical energy and vice versa, so acting as a both a transmitter and receiver.

WAVE TYPES

Continuous wave (CW)

Applying an alternating current (AC) voltage causes continuous expansion and contraction at same frequency. Conversely, applying an alternating pressure causes an alternating voltage at the same frequency. This wave takes a sinusoidal form (Fig. 7.1A and B).

Pulsed wave (PW)

Applying a direct current (DC) voltage causes a short period of expansion and compression, due to the elasticity of the medium. After the DC 'hit', the transducer continues to vibrate, but loses energy exponentially (in the form of sound), producing a short pulse of ultrasound (Fig. 7.5). This type of pulse is used in ultrasound imaging. Pulse length can be defined in terms of time or distance. The length of an imaging pulse is typically up to 3 wavelengths or periods. For example, at 3 MHz:

$$\text{Pulse Length (distance)} = 3\lambda = 3 \times \frac{c}{f}$$

$$= 3 \times \frac{1540}{3 \times 10^6} = 1.5 \text{ mm}$$

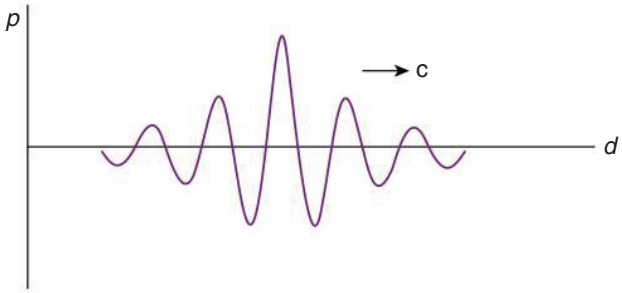

Fig. 7.5 Excess pressure versus distance for a pulsed wave.

$$\text{Pulse Length (time)} = 3T = 3 \times \frac{\lambda}{c}$$

$$= 3 \times \frac{1.5 \times 10^{-3}}{1540} = 1\ \mu\text{s}$$

As speed is assumed to be the same in tissue, time and distance can effectively be interchanged.

Damping, when applied to an ultrasound transducer, causes loss (decay) of energy and shortens the pulse length (Fig. 7.6). **Ringing** is the continuation of pulse vibrations and occurs when there is little or no damping.

Frequency spectrum

In CW, a single frequency (e.g., a pure tone) is emitted. The frequency spectrum, which plots relative intensity against frequency, is a single line A in Fig. 7.7. In PW, the pulse contains a range of frequencies: a continuous spectrum of sine waves that combine to produce the pulse B or C is shown in Fig. 7.7.

Bandwidth describes the range of frequencies contained in a pulse and is defined as the full width at half maximum of the frequency spectrum. The shorter the pulse, the larger the bandwidth.

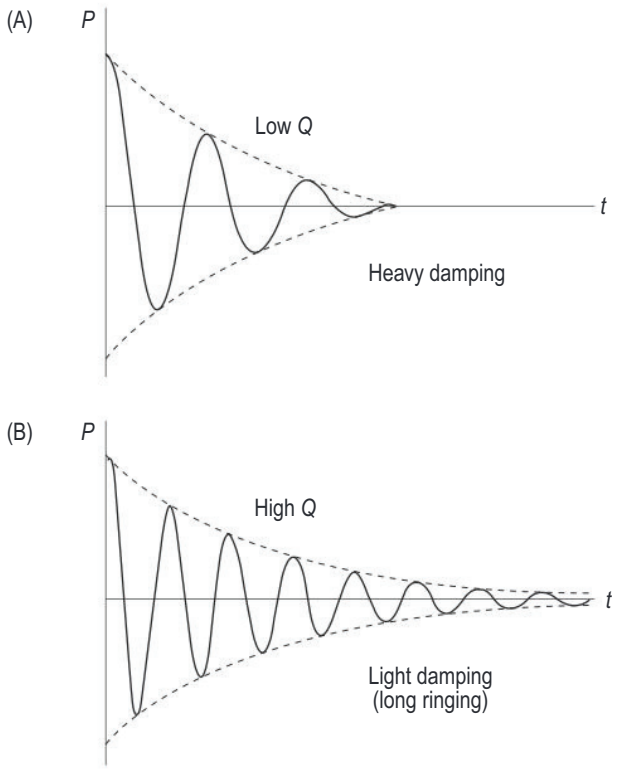

Fig. 7.6 Damping: (A) heavy and (B) light.

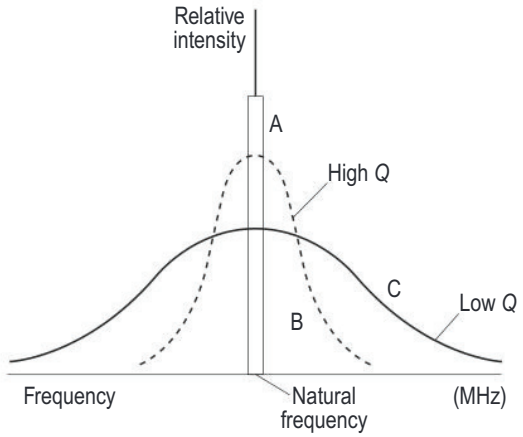

Fig. 7.7 Frequency spectrum in (A) continuous and (B and C) pulsed modes

SINGLE-TRANSDUCER PROBE

Fig. 7.8 shows the basic components of a single-transducer ultrasound probe. The piezoelectric element is used to transmit and receive the ultrasound.

Resonant (natural) frequency

In Fig. 7.9, the front face of the transducer emits sound both forwards and backwards. The back wave *B* is reflected at the back face. By the time it meets the front wave F, it has travelled an extra distance of twice the crystal thickness: 2*t*. If, as in the diagram, this distance is equal to one

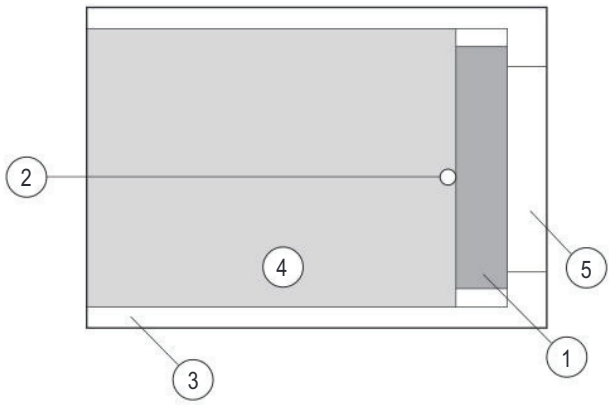

Fig. 7.8 Section through a single-transducer probe. *1*, Piezoelectric element; *2*, insulated wire; *3*, earthed metal case; *4*, backing block; *5*, lens.

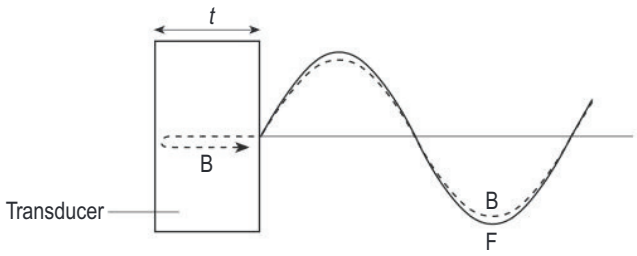

Fig. 7.9 Resonance. *B* and *F* are the backward and forward-travelling waves, respectively. *t* is the transducer crystal thickness.

wavelength. *F* and *B* are perfectly in phase, resulting in maximum constructive interference. The corresponding frequency is the resonant frequency and therefore depends on *t* and the speed of sound in the crystal. If using DC, the transducer will naturally vibrate at the resonant frequency, with a wavelength of:

$$\lambda = 2t$$

For example: for a 4 MHz transducer in PZT crystal, the thickness would be:

$$\frac{1}{2}\lambda = \frac{1}{2} \times \frac{c}{f} = \frac{1}{2} \times \frac{4000}{4 \times 10^6} \approx 0.5 \text{ mm}$$

Hence, a thicker crystal resonates at a lower frequency, producing a longer wavelength.

In general, a range of thicknesses can be used when driven with AC at a chosen resonant frequency (where *n* is any integer):

$$\lambda = 2n \times \textit{crystal thickness}$$

The transducer is most sensitive when operating at the resonant frequency. Most modern transducers operate across a bandwidth (e.g., 1–5 MHz). To use a frequency outside this range, one would have to change transducer.

Mechanical coefficient (Q factor) describes the bandwidth of frequencies produced by a given crystal (Fig. 7.7), defined by the centre frequency (f_0) and the bandwidth (Δf):

$$Q = \frac{f_0}{\Delta f}$$

A high Q transducer produces a pure tone and only responds to a single frequency, as in CW. A low Q transducer has a short ring-down time, produces short pulses, and produces and responds to a wide range of frequencies, as in PW.

Electrodes

The voltage is applied to the crystal between an insulated wire and earthed metal case.

Backing Block

The transducer is mounted onto a block of material such as tungsten powder in epoxy resin. This is *matched* to the transducer material to enable transmission backwards and is used to absorb the ultrasound and damp the vibration, like placing a hand on a vibrating drum skin, enabling shorter pulses. The disadvantages of damping are reduction in output and reduced Q factor. CW ultrasound transducers are therefore typically air-backed.

Lens

A layer of material (the lens) such as plastic or silicone is affixed to the front face of the transducer, performing several functions:

- Protects the surface of the element from damage
- Insulates the patient from the applied voltage
- Focuses the beam (like an optical lens)
- Improves transmission into the patient from the transducer by acoustic 'matching'

ULTRASOUND BEAM

If the element diameter (aperture size) D is smaller than one wavelength (e.g., 0.5 mm), sound will spread equally in all directions as spherical waves and would have no directional properties (Fig. 7.10A).

If D is much larger, sound is projected forwards, effectively as a plane wave. As in Fig. 7.10B, the transducer can be thought of as several mini-transducers. For every crest that reaches any point B outside the beam from one mini-transducer, a trough arrives from another, causing destructive interference and in general cancelling out. Within the beam, the separate sound waves that reach point A within the beam are more or less on phase, so constructively interfere and reinforce. As a result, most of the energy is confined within an ultrasound beam of width D.

Focus: The intensity of the beam is naturally concentrated at a defined depth, known as the beam focus. At this depth, the amplitude of the beam is increased and the width of the beam at its narrowest. The behaviour of the ultrasound beam changes with depth and can be divided into two portions, divided by the beam's natural focus:

The **near field (Fresnel region)** extends from the transducer face and remains nearly parallel up to the focus. The near field length is calculated by:

$$N = \frac{\left(D/2\right)^2}{\lambda}$$

Hence, N is proportional to fD^2.

The **far field (Fraunhofer region)** extends beyond the focus, where the interference effect is lost, and the beam diverges. The angle of divergence is calculated by:

$$\sin\theta = 1.2\frac{\lambda}{D}$$

Hence, increasing frequency or aperture size reduces far field divergence.

For example: at 3.5 MHz with aperture size $D = 12$ mm:

$$N = \frac{\left(D/2\right)^2}{\lambda} = \frac{\left(6 \times 10^{-3}\right)^2}{4.4 \times 10^{-4}} = 82 \text{ mm}$$

$$\theta = \sin^{-1}\left(1.2\frac{4.4 \times 10^{-4}}{12 \times 10^{-3}}\right) = \sin^{-1}(0.044) = 2.5^\circ$$

BEHAVIOUR OF A BEAM AT INTERFACES

If a beam strikes the boundary between two media (transducer-skin, tissue-bone, tissue-air etc.), some energy is reflected as an 'echo' and some is transmitted through the interface.

The portion of the beam that is reflected towards the transducer is detected and contributes to the image. The portion that is transmitted continues through the tissue and is either detected due to subsequent reflections or is lost due to attenuation.

Specular reflection

This occurs if:

- The beam strikes a large, smooth interface that is *larger than the wavelength*
- The acoustic impedance of the tissues on either side of the interface are non-equal ($Z_1 \neq Z_2$)

It is analogous to light reflecting off a mirror. As with light, the angle of reflection is equal to the angle of

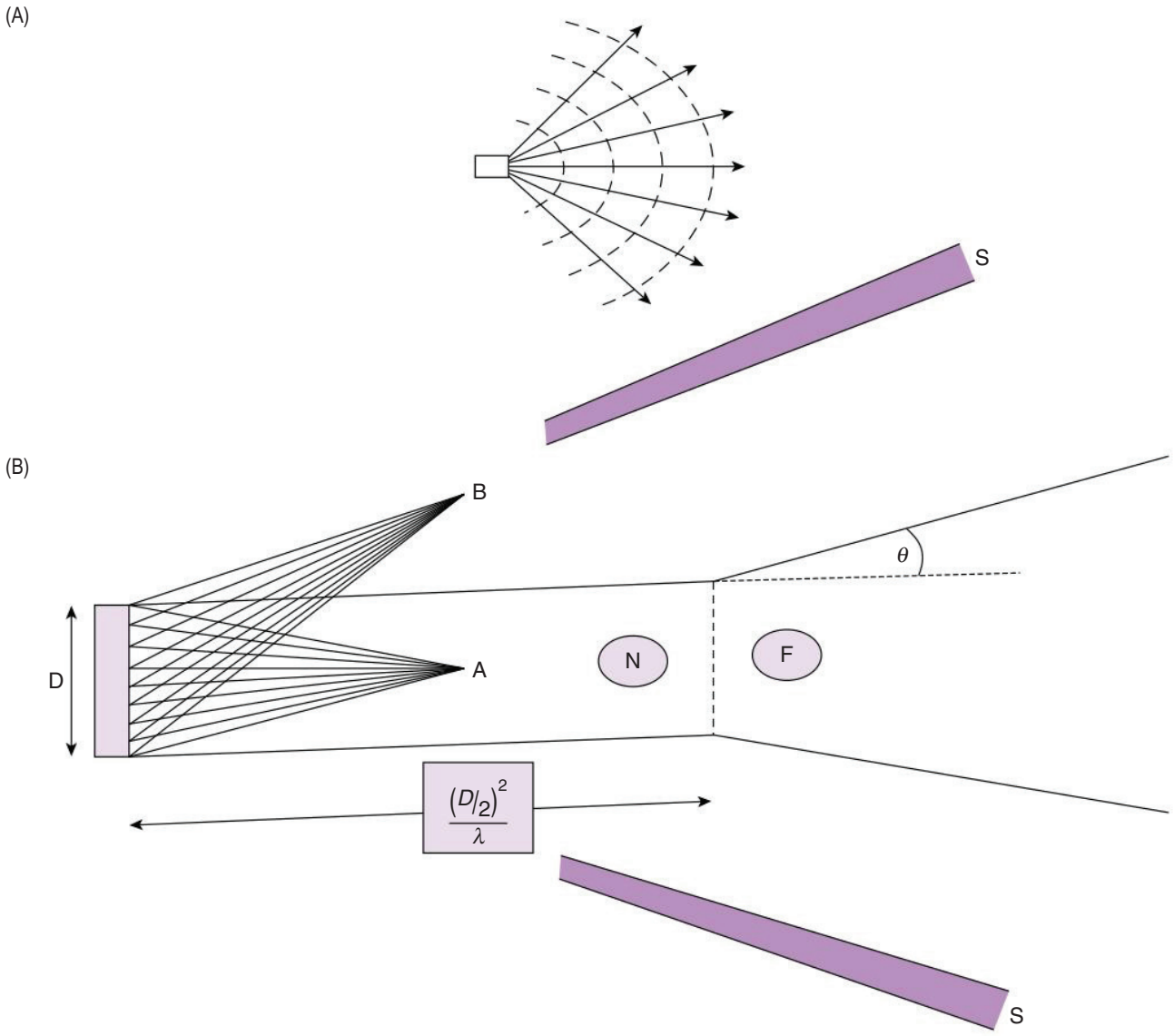

Fig. 7.10 Pattern of sound emitted by (A) a very small aperture: dashed lines represent wavefronts and solid lines show direction of propagation; and (B) a larger aperture with near (N) and far (F) fields.

incidence (Fig. 7.11A). If the ultrasound beam is at or close to right angles to the interface, the fraction of the beam that is reflected *(R)* can be calculated as:

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

Hence, the greater the difference in *Z*, the greater the fraction *R* reflected (Table 7.3).

Small variations in soft tissues result in small fractions of reflection, e.g., nearly 1% at a kidney-fat interface. This enables visualisation of internal anatomical structures such as organ boundaries.

For example: for a bone/muscle interface, the reflected fraction is:

$$R = \frac{(5.3 - 1.7)^2}{(5.3 + 1.7)^2} \approx 30\%$$

Hence, approximately 70% is transmitted through the interface. In general, it is not possible to image through bone due in part to reflection, but also to high levels of absorption.

In air, *Z* is negligible, resulting in total reflection and no transmission. Anatomy that is behind gas-filled organs cannot be imaged. The bowel wall can therefore be imaged,

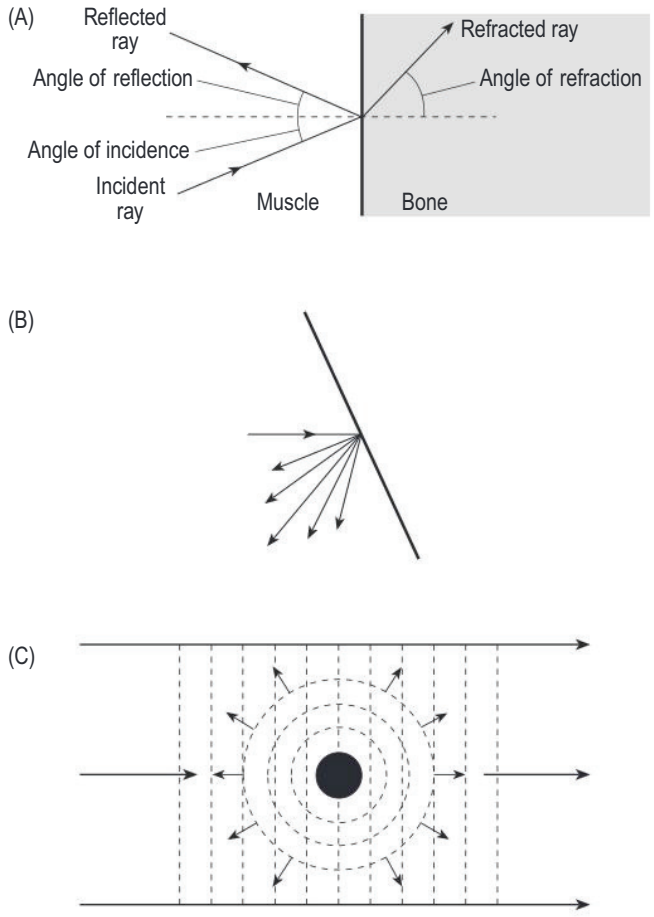

Fig. 7.11 (A) Specular reflection, transmission, and refraction. The angle of reflection is equal to the angle of incidence and the angle of refraction is determined by Snell's law. (B) Diffuse reflection and (C) scattering.

but not the lumen itself. Sound cannot travel from the transducer into the tissue if air is trapped between the transducer face and the skin. Hence, acoustic coupling gel is used, and bubbles must be avoided.

Acoustic matching

If $Z_1 = Z_2$:

$$R = 0$$

$$\text{therefore } (1 - R) = 1$$

Hence, 100% of the energy is transmitted. This means that the media are *acoustically matched*, as should be the case for the transducer and backing block.

Matching layer. PZT crystal has a very high Z (Table 7.1), so if transmitting directly into tissue, most of the ultrasound is reflected:

TABLE 7.3 Typical Reflection Factors

Interface	Reflected Fraction (%)
Gas-tissue	99.9
Soft tissue – lead zirconate titanate	80
Bone-muscle	30
Plastic–soft tissue	10
Fat-muscle	1
Blood-muscle	0.1
Liver-muscle	0.01

$$R = \frac{(30 - 1.5)^2}{(30 + 1.5)^2} \approx 80\%$$

To allow greater transmission, a matching layer is used (often in the form of the lens), with an intermediate acoustic impedance Z_2 chosen such that:

$$Z_2 = \sqrt{Z_1 Z_3}$$

For the above example:

$$Z_2 = \sqrt{30 \times 1.5} = 6.7$$

The thickness of the matching layer also affects transmission, due to interference.

Diffuse reflection

This is the same effect as that seen with light and frosted glass. It occurs if the beam strikes a rough interface with *undulations approximately equal to 1 wavelength*. The incident beam is reflected over a range of angles (Fig. 7.11B). This spread is greater on rougher surfaces and for shorter wavelengths. When this happens at multiple surfaces, some reflections may return to the transducer even if they did not strike an interface at right angles. Diffuse reflections contribute to most of the generated ultrasound image.

(Rayleigh-like) scatter

This occurs if the beam encounters a structure that is *much smaller than the wavelength* (e.g., red blood corpuscle, diameter 10 μm, or tissue parenchyma). The beam is scattered in all directions and produces an interference

pattern that does not directly represent but is related to the tissue structure, as well as the wavelength (Fig. 7.11C). This enables visualisation and characterisation of the interior of tissues such as placenta, liver, pancreas, spleen, and kidney. The signal received from scatter is usually 1%—10% as strong as from organ boundaries.

Refraction

This occurs at an interface if the speeds of sound of the tissues on either side of the interface are non-equal ($c_1 \neq c_2$). The portion of the beam that is transmitted will continue along an altered angle, θ_2 (Fig. 7.11A), calculated using Snell's law:

$$\frac{\sin\theta_1}{\sin\theta_2} = \frac{c_1}{c_2}$$

ATTENUATION

When travelling through a medium, sound energy is lost to the surrounding medium. This happens due to:

- **Absorption** — As the ultrasound travels through tissue, processes such as friction and viscous forces convert the mechanical energy into heat energy, which is absorbed by the tissue. In soft tissue, absorption is approximately proportional to frequency, so higher frequencies are more readily absorbed.
- **Scattering** and **reflection** from interfaces, removing energy from the forward-travelling beam.

Attenuation is quantified using the ratio of the power or intensity of the incident and transmitted beam — P_i and P_t (W) *or* I_i and I_t (W/m^2 or W m^{-2}) — over a given distance x. As ratio values can vary widely, it is simpler to represent these on a logarithmic (rather than linear) scale in decibels (dB).

$$\textit{Decibels} = 10\log_{10}(\textit{Power or Intensity Ratio})$$

Note that decibels are additive. Positive values show amplification and negative values attenuation. The attenuation of a medium is quantified by the attenuation coefficient as follows:

$$\textit{Attenuation Coefficient} = -\frac{1}{x}10\log_{10}\left(\frac{I_t}{I_i}\right)\text{dB cm}^{-1}$$

Typical values are given in Table 7.4.

There is little attenuation in water, so a full bladder can allow the ultrasound to reach deeper structures. Bone and air attenuate much more, so it is very difficult to image through ribs and bowel gas.

The **half-value layer** is the thickness of tissue that reduces the intensity to half. This corresponds to a change of: $\log_{10}\left(\frac{1}{2}\right) = -3$ dB. For example: the half-value layer at 1 MHz in average tissue (0.7 dB/cm) would be:

$$HVL = -\frac{\log_{10}\left(\frac{1}{2}\right)}{0.7} = 4.3\text{ cm}$$

At a certain depth, the intensity of the beam has reduced so much that it is no longer useful. This depth of penetration is poorer at higher frequencies, as attenuation is greater. Roughly, penetration (cm) $= {}^{40}/_{f}$.

A-MODE (AMPLITUDE MODE)

A-mode is the simplest form of ultrasound imaging, producing a simple representation of the depth of tissue interfaces along a single line (Fig. 7.12). This has now been largely replaced by B-mode, but has been used for examining the eye, identifying midline displacement in the brain, and identifying cysts in the breast.

1. The probe face is held against the patient and an ultrasound pulse is transmitted into the tissue.
2. This pulse takes a time t to reach interface *a*.
3. Some of the energy is reflected back along the path towards the transducer, taking an additional time t to reach the transducer (i.e., a total round-trip time of $2t$).

TABLE 7.4 Typical Tissue Attenuation Values (Approximate)

		HALF-VALUE LAYER (CM)		
Tissue Type	**Attenuation (dB/cm at 1 MHz)**	**1 MHz**	**2 MHz**	**5 MHz**
Water	0.0022	1360	340	54
Blood	0.18	17	8.5	3
Average Tissue	0.7	4.3	2.1	0.9
Bone	15	0.2	0.1	0.04
Lung	40	0.08	0.04	0.02

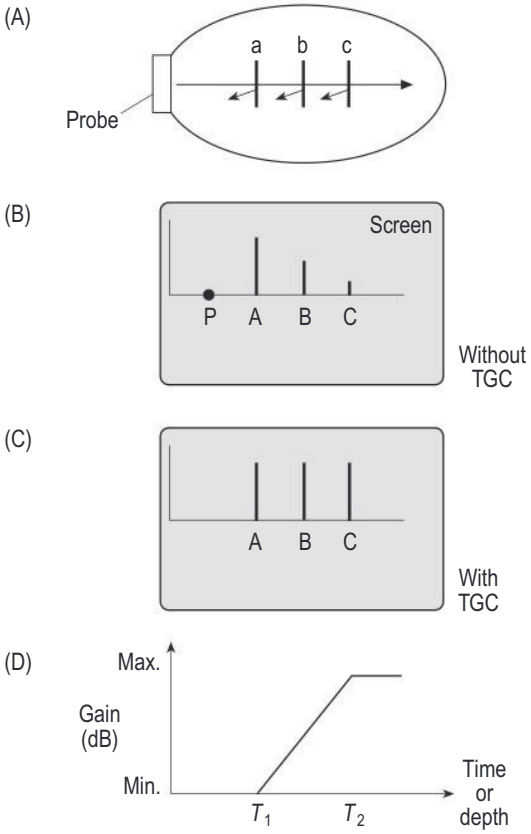

Fig. 7.12 A-mode: (A) section through transducer and patient, (B) trace on screen without time gain control, (C) trace on screen using time gain control, and (D) variation of gain with depth. *TGC*, time gain compensation.

4. The transducer acts as a receiver and converts the sound energy into an electrical pulse.
   The depth of the interface can be calculated using:

$$depth_a = c \times t_a$$

   where c is the assumed speed of sound and t = half the time between transmission and reception.

5. This signal is amplified and a short vertical trace ('blip') is produced on the screen at the calculated depth. The other interfaces *b* and *c* then produce corresponding blips.
6. Pulses are transmitted and received at regular intervals to refresh the image.

B-MODE (BRIGHTNESS MODE)

Rather than imaging a one-dimensional line through the patient, B-mode allows imaging of a two-dimensional slice (similar to a single slice in computed tomography).

As in A-mode, the transducer is pulsed at regular intervals, however in B-mode, the ultrasound beam is scanned sequentially across a two-dimensional section, either:

- Linearly to produce a rectangular image
- Rotationally to produce a sector image

Fig. 7.13 shows the scan lines travelled by each pulse to produce the image. Only boundaries approximately perpendicular to the scan lines will be imaged.

The returning echo pulses are displayed as bright pixels on the screen. The image is built up as a series of vertical lines on the screen, corresponding to each ultrasound beam in the sequence, and displayed on the monitor screen in a matrix of pixels (e.g., 512 × 512 or 1074 × 768), corresponding to the matrix of voxels in the scanned anatomy:

- The lateral (horizontal) pixel position is determined by the position of the beam relative to the patient surface, or the angle of the beam.
- The axial (vertical) pixel position is determined by the depth calculated from the round-trip time.
- The brightness of the pixel is determined by the strength of the received signal. As the signal strength depends on the strength of reflection from the corresponding interface, this enables tissue differentiation based on the pixel brightness.
- A real-time image is displayed on the monitor screen in a matrix of pixels, corresponding to the matrix of voxels in the body.

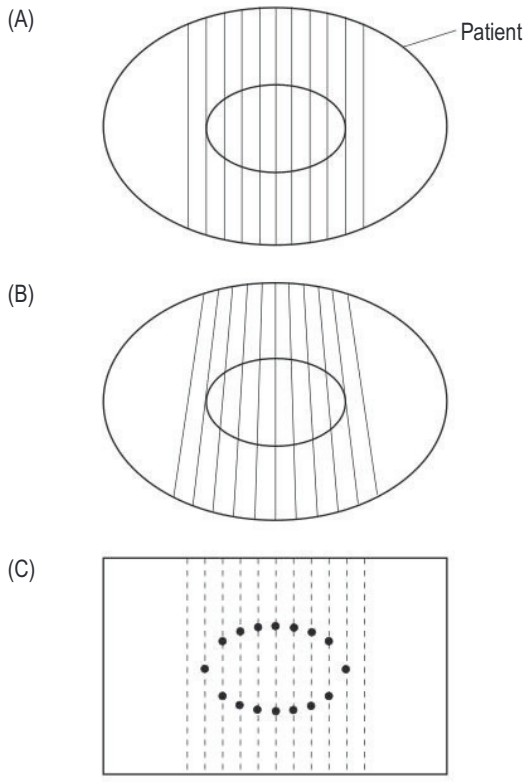

Fig. 7.13 B-mode: (A) linear scan, (B) sector scan, and (C) the monitor screen.

TABLE 7.5 Methods of Producing a Two-Dimensional Image

Scan Method	Pros	Cons	Applications
Mechanical Sector Scanning	• Cheap • Focussed beam allows better resolution than phased array	• Moving parts • Probes can be cumbersome	• Transrectal imaging • 4D imaging
Electronic Linear or Curvilinear Array	• Wide field throughout the image • Good image quality	• Wide probe face requires a large area of patient contact • Curvilinear line density reduced with depth due to beam divergence	• Flat linear probes: superficial vessels and nerves, thyroid • Curvilinear probes: abdomen, liver, obstetrics and gynaecology
Electronic Steered/Phased Array	• Smaller probe face so requires a smaller acoustic window • Smaller probe is easier to manipulate • Wider field at depth	• Narrower field at the surface • Generally lower image quality	• Used to image the heart through intercostal spaces or the infant brain through the fontanel • Also used for intracavity probes, including endoscopic probes for imaging the heart

Ultrasound images can be produced in a variety of ways depending on the clinical applications and each has its own advantages and disadvantages (Table 7.5).

REAL-TIME IMAGING

In most B-mode scanners, these two-dimensional images are produced continuously in a rapid succession of frames so that moving structures can effectively be viewed in real-time. Real-time imaging also allows large volumes to be scanned in a short amount of time. Various aspects of the real-time image can be influenced directly or indirectly by the user either by choice of transducer, or of scanner settings.

Scan line density:

The image is divided into several vertical lines with a width defined by the distance between each beam. The greater the number of lines per unit distance, or line density, the better the lateral resolution.

Pulse repetition frequency (PRF):

The rate at which pulses are transmitted along one line, measured in Hz (pulses per second):

$$PRF = \frac{1}{time\ between\ pulses}$$

Depth of view:

To image structures at depth, a pulse must have time to make the full round trip to and from the deepest structure, so:

$$depth\ of\ view = 0.5 \times \frac{c}{PRF}$$

Frame rate:

To produce a two-dimensional frame, all lines in the frame must be produced sequentially, before starting again to produce the next frame. The frame rate is the number of frames produced each second (Hz).

$$frame\ rate = \frac{PRF}{lines\ per\ frame}$$

Hence, frame rate can be increased by increasing PRF or reducing line density.

For example: for an image with 100 lines, to achieve a frame rate of 30 Hz would require:

$$PRF = frame\ rate \times lines\ per\ frame = 30 \times 100 = 3\ kHz$$

To successfully image moving structures, a sufficiently high frame rate is required. If the frame rate is too low, this can result in image 'lag' and blurring, particularly as the probe is moved across the patient.

It is therefore not possible to achieve a high frame rate, high line density, and image at a large depth, and these aspects must be balanced. Combining the above:

$$\textit{depth of view} \times \textit{number of scan lines} \times \textit{frame rate} = \textit{constant}$$

SCAN TYPES

Two-dimensional B-mode images can be achieved in a number of ways, each with their own advantages (Table 7.5).

Mechanical (Sector) Scanning

This technique uses a transducer which is moved inside a fluid-filled outer case that is pressed against or inserted into the patient. This can be oscillated back and forth by an electric motor (Fig. 7.14), or a small group of transducer elements may be mounted onto a rotating rod (Fig. 7.15).

Electronic Scanning: Stepped Linear or Curvilinear Array

The more common technique for producing a scanned image is to use an elongated transducer that is divided up into multiple narrow transducer elements (typically at least 128) (Fig. 7.16).

As each element is very narrow, individually these would produce a poor beam for imaging, with a short near field and widely diverging far field. To mitigate this effect, elements are energised in overlapping groups in succession, (e.g., 1–6, 2–7, 3–8) so that a well-defined ultrasound beam can be scanned across a rectangular area with (say) 512 scan lines.

For this type of **linear transducer**, the size of the field of view is limited by the width of the probe (Fig. 7.17). Alternatively, in a **curvilinear transducer**, the transducer elements can be arranged on a curved surface (Fig. 7.18). As the lines of sight are perpendicular to the transducer surface, they spread out and can cover a wider field of view (Fig. 7.19).

Electronic Sector Scanner: Steered/Phased Array

This transducer has the same design as a stepped linear array, but is shorter, containing fewer elements. Instead of energising small groups of elements, a phased array uses all elements.

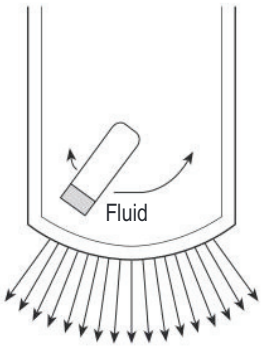

Fig. 7.14 Mechanical sector scanner with an oscillating transducer.

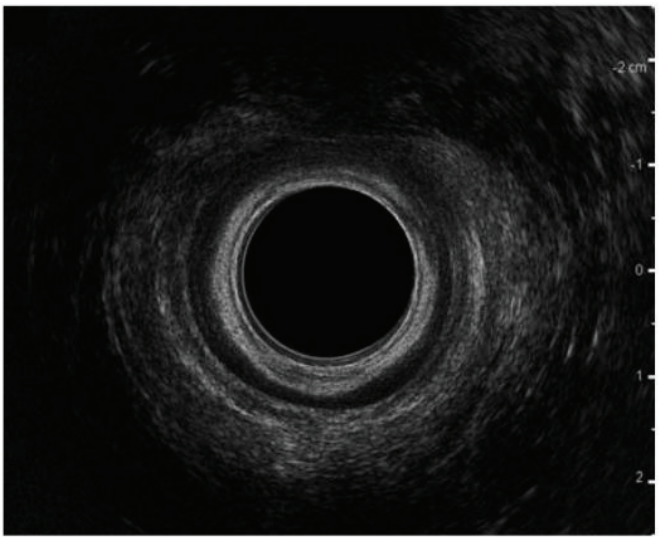

Fig. 7.15 Image of the rectum wall produced using a mechanical rotating transducer mounted inside a rod-shaped endorectal probe.

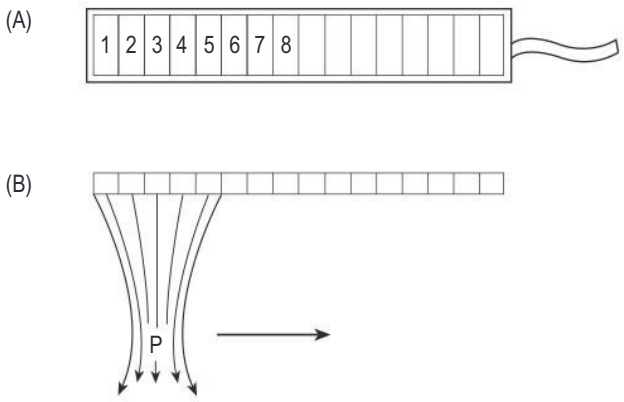

Fig. 7.16 Linear transducer array: (A) transducer face and (B) cross-section through transducer and ultrasound with electronic focussing across the scan plane.

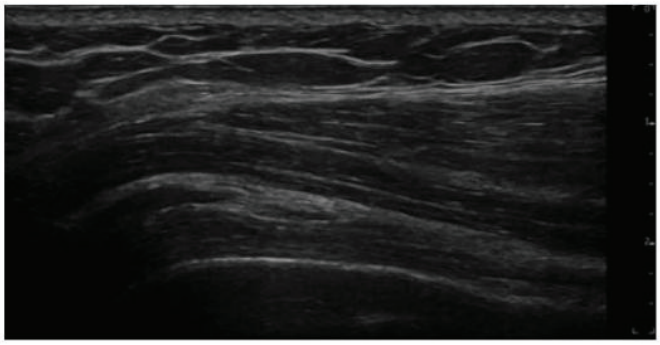

Fig. 7.17 Image of rotator cuff muscles using a linear array.

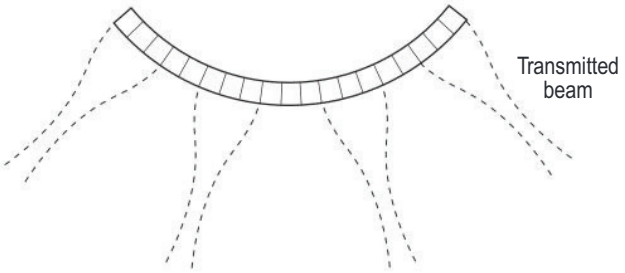

Fig. 7.18 Curvilinear array with beams spreading out perpendicular to the curved probe surface.

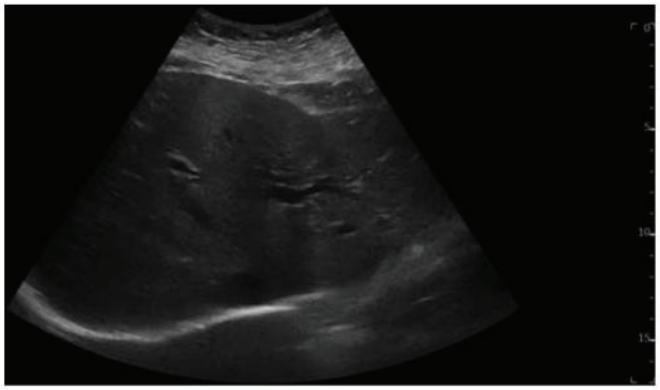

Fig. 7.19 Sector-shaped image of liver using a curvilinear transducer.

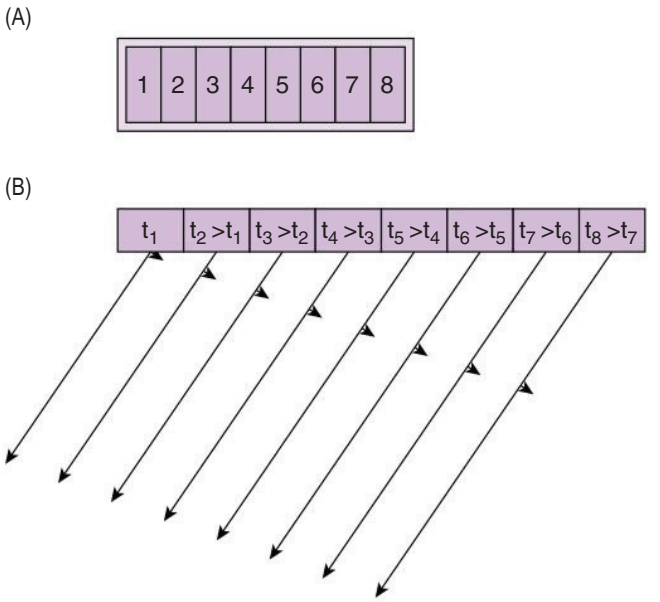

Fig. 7.20 (A) Phased array (B) timings and resulting steered beam.

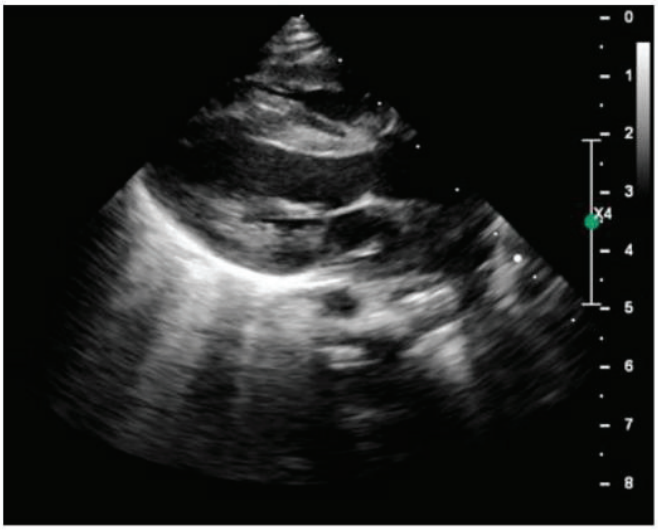

Fig. 7.21 Image of the paediatric heart using a phased array.

Steering: If energised simultaneously, the individual elements act as a single transducer with a beam directed forwards. If small delays are introduced in rapid sequence along the transducer, the pulses reinforce at an angle from the transducer face and destructively interfere in all other directions, producing a steered plane wave (Fig. 7.20). Changing the delay timing changes the angle of the beam, so this can be swept across the field of view producing a sector-shaped image (Fig. 7.21).

Intracavity Probes

To reduce the need for good penetration and to avoid the obscuring effects of bone or bowel gas, transducer arrays can be arranged on a probe that is designed for imaging intracavitarily (Table 7.6).

TABLE 7.6 Typical Types of Intracavity Probes

Probe Type	Application	Variations
Transrectal	• Prostate • Rectum wall	• Linear array along the length • Tightly curved array at the end for a sector view • Rotating mechanical probe
Transvaginal	• Gynaecology • Obstetrics	• Tightly curved array at the end for a sector view
Endoscopic	• Cardiac (transoesophageal) • Transvascular	• Small, high-frequency phased array • Very small array with miniaturised electronics

POWER AND GAIN

The amplitude of the returning echo depends on the amplitude of the incident beam, the ratio of acoustic impedances, and the level of attenuation from interceding tissues. Hence, some interfaces may not be visualised at lower transmission amplitudes. Power and gain are two strategies for overcoming this:

Power: Increasing the output power increases the amplitude of the transmitted wave so that greater amplitude reflections are received. This also imparts more energy into the tissue, contributing to greater tissue heating.

Gain: If small signals are still detected by the transducer but not seen clearly on the image, the brightness can be increased by electrical amplification of the received signals, or increasing the overall gain. This amplifies both the 'real' received signal and any electrical noise inherent to the system.

Time gain compensation: Due to attenuation, the transmitted ultrasound energy diminishes with increasing depth, as does the returning echo, so that interfaces deeper in the body produce weaker echoes. Attenuation is compensated for by using time gain compensation (TGC), which increases the amplification of the received echoes as time increases. The aim is to render all echoes from identical interfaces the same, independent of their depth.

In A-mode, the user can alter the slope of the ramp and (retain indent) the resulting TGC curve (Fig. 7.12). In B-mode, the user has more freedom. The image is divided into more than a single range of depths and the user can adjust the level of amplification for each depth range to achieve an optimised image.

FOCUSSING

The beam width is narrowest in the near field and at the natural focus, but this can be further reduced using additional focussing. Stronger focussing leads to a shorter focal length and a narrower (and shorter) focal region, but also leads to greater beam divergence beyond the focus. Note that the focal length for a focussed beam is not necessarily the same as the length of the near field.

Physical Focussing

Physical focussing can be achieved in a few ways (Fig. 7.22):

- A curved piezoelectric element achieves a shorter focal length using greater curvature
- An annular array uses multiple elements arranged concentrically
- A plastic or silicone acoustic lens moulded to the front of the transducer (can be convex or concave, depending on material)

Electronic Focussing

In addition to physical methods, the beam can be focussed by introducing delays in when each element in a group is energised, using a similar technique to steering. This allows the user to choose the depth at which the lateral resolution of the image should be optimised.

- The outermost pair is energised first, then each adjacent pair in succession, ending with the centre (Fig. 7.23).
- The focal point, P, is defined by the depth at which the pulses arrive together and reinforce: the greater the delays, the shorter (shallower) the focal point.

Focussing improves the beam width at the focal depth but also causes greater divergence beyond the focus. This can be mitigated using multiple-zone focussing:

- For each line, multiple pulses are sent with different focal depths.
- The transducer is gated so it receives only echoes from the chosen focal depth.
- The data from each focal depth is then stitched together to produce one frame with several focal depths.

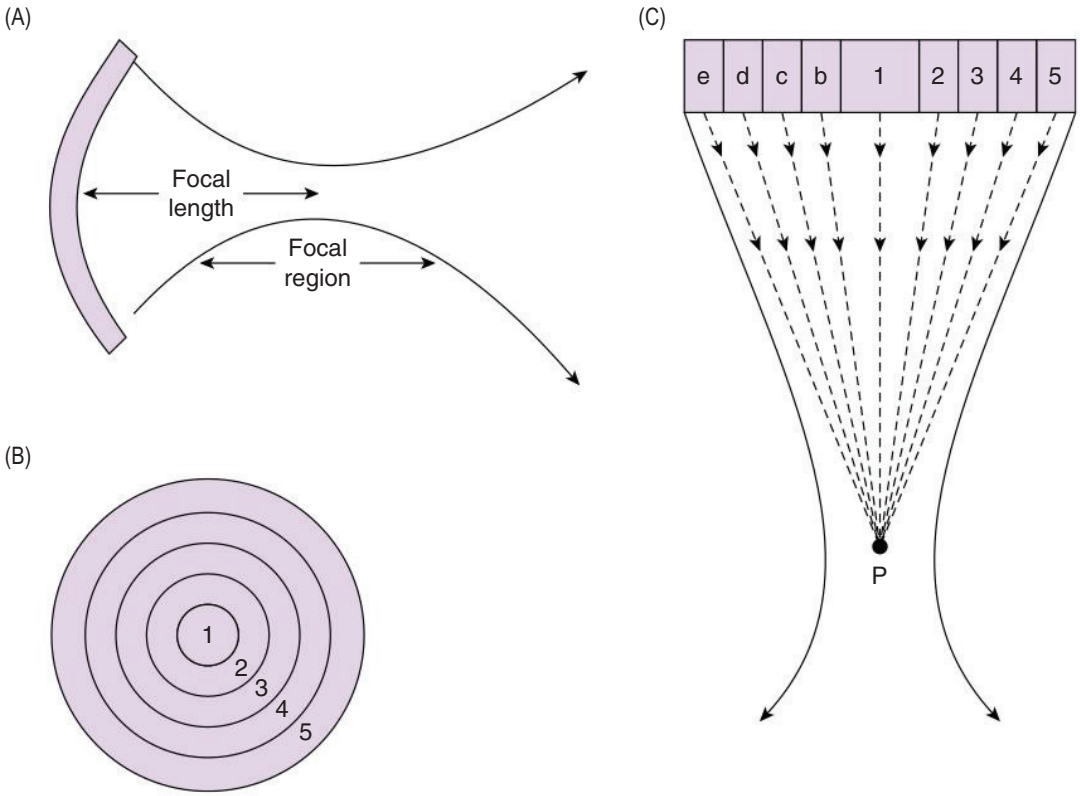

Fig. 7.22 (A) Focussed beam from a curved transducer, annular array, (B) transducer face, and (C) cross-section through the transducer and resultant focussed beam.

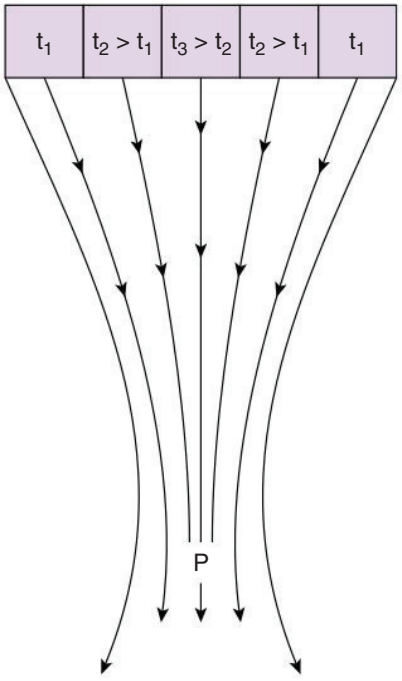

Fig. 7.23 Electronic focussing.

- If the focussing is done in transmission, multiple sets of pulses are required, so the frame rate reduces. Alternatively, instead of introducing delays to the transmitted beam, these can be added after reception. The result is that multiple focal depths can be created using the same transmitted beam and frame rate is maintained.

Note that this type of focussing is in only one plane: the azimuthal, or scan plane, parallel to the array. Focussing in the perpendicular or elevation plane is usually done using physical methods and this defines the slice thickness. Electronic focussing in the slice plane can be achieved using a '1.5D transducer', using multiple layers rather than a single layer of elements to improve the slice resolution.

TISSUE OPTIMISATION

In standard ultrasound imaging, the speed of sound is assumed to be 1540 m/s, when in fact in soft tissue this can range between 1300 and 1800 m/s. Fatty tissues such as breast tissue fit into the lower end of this range, so that aspects of image production such as electronic focussing

timings are inaccurate, leading to reduced spatial resolution increased clutter.

The speed of sound can be corrected by the user, or **tissue optimisation** can be applied automatically by the system, through analysis and optimisation of features in the image such as spatial frequencies, which relate to spatial resolution.

M-MODE

M-mode is used to examine the position of moving structures over time.

- The B-mode image is frozen and used to direct a single beam (as in A-mode) along a line of interest, intersecting the moving surfaces as close to right angles as possible (Fig. 7.24).
- This line of echoes is displayed vertically on the screen.
- Each new line is displayed alongside the last one to show how the position changes with time horizontally across the screen.

The high temporal resolution of this technique allows quantitative analysis of fast-moving structures such as heart valves and the heart wall, which is hard to achieve with B-mode.

THREE- AND FOUR-DIMENSIONAL IMAGING

Three-dimensional images are obtained from a set of two-dimensional scans which are reconstructed to produce a volumetric image. This can be done in the following ways:

- The transducer is swept manually across the patient skin. Each frame is then stitched together. The geometric accuracy therefore depends on the skill of the operator.
- The transducer is moved mechanically within the probe. This requires a bulky probe with internal moving parts.
- A two-dimensional array is used to acquire two-dimensional images in two orthogonal planes, enabling real-time three-dimensional imaging, known as four-dimensional imaging. Many scanners will also perform post-processing such as volume and surface rendering

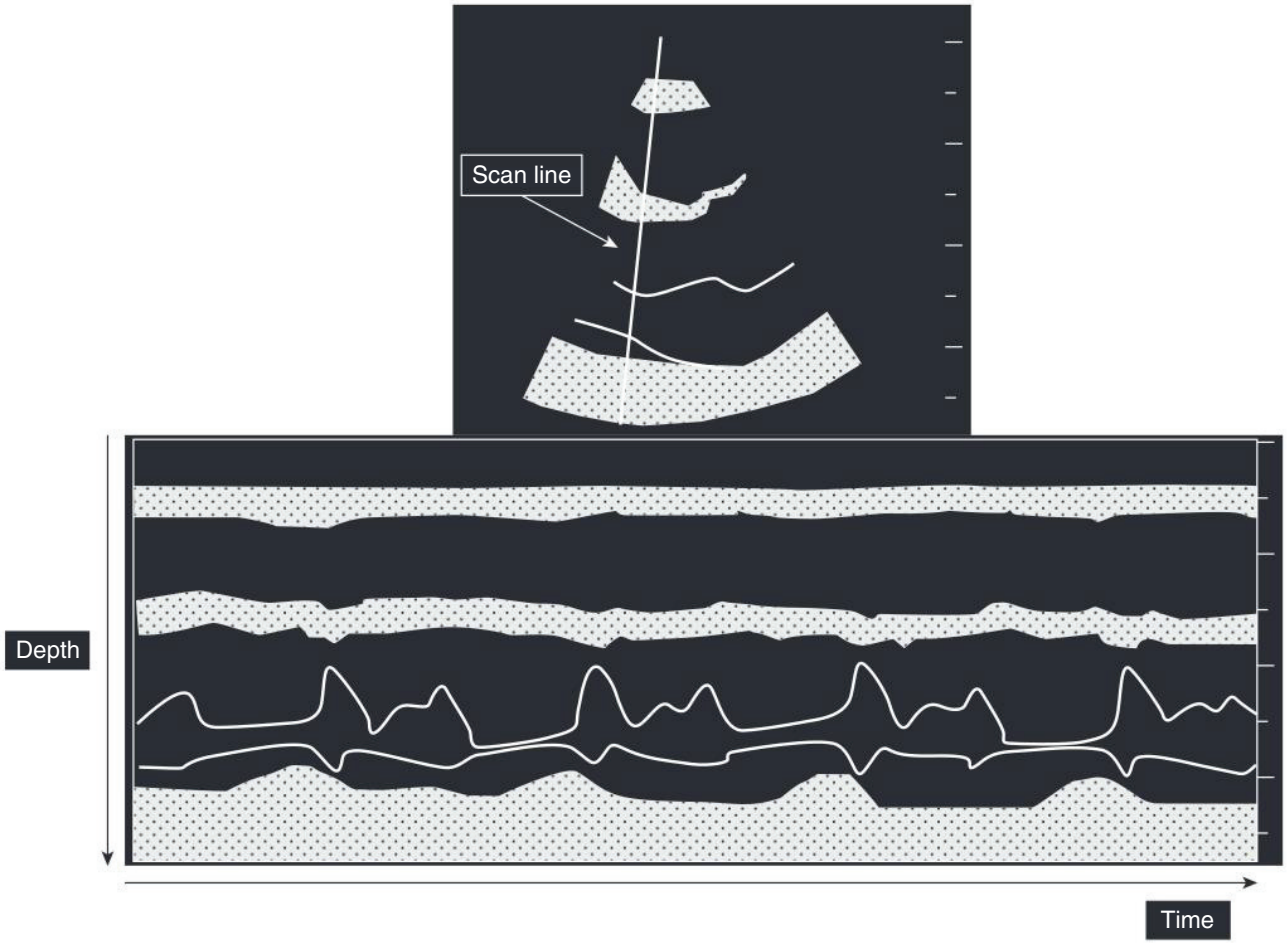

Fig. 7.24 M-mode.

to produce a three-dimensional image. This can provide useful information regarding vasculature and for foetal imaging.

IMAGE QUALITY

Signal-to-Noise Ratio (SNR)

Noise is any spurious information in the image that does not correspond to real structures in the patient. In ultrasound, noise is principally electronic noise: statistical fluctuations in the number of electrons in the very small currents measured. This appears on the image as a randomly changing 'salt-and-pepper' haze. Noise can also include any other artefacts that may obscure real anatomy, sometimes known as 'clutter'.

The **SNR** is the ratio of real signal to noise and greater SNR corresponds to better visibility of structures.

$$SNR = \frac{Real\ Signal}{Noise}$$

Contrast

The contrast between two structures is determined by the difference in the amplitudes of the reflections from each structure. High contrast makes it easy to distinguish between two structures.

Spatial Resolution

Axial (or depth) **resolution** is the ability to separate two interfaces along the same scan line. If the interfaces are too close together, the echo pulses will overlap and be recorded as a single interface (Fig. 7.25).

Axial resolution is about half the pulse length, so shorter pulses produce better axial resolution. This can be achieved by:

- Greater damping (i.e., a low Q) achieved through use of a backing layer
- Increased frequency: keeping the number of cycles in a pulse constant, a shorter wavelength will result in a shorter pulse

Lateral resolution is the ability to separate two structures side-by-side at the same depth. This depends on the beam width being narrower than the gap.

In Fig. 7.26, the width of the beam is smaller than the gap between structures A and B, so structure A is only intersected by beam 1 and structure B by beam 3. Where the beam diverges, it is wider than the gap between structures C and D, meaning that both structures are intersected by all 3 beams. Echoes from the edges of the beam are therefore misregistered in the image as having originated from the centre of the beam.

Poor lateral resolution leads to a *smearing* of small details and edges across the image. Lateral resolution in the near field is improved by using a smaller aperture and focussing.

In the focal region of the beam, axial resolution may be about one wavelength and lateral resolution about 3 times worse, being about one-third of the aperture diameter.

Slice resolution is the ability to image a structure in a narrow plane, without contribution from structures in adjacent planes. This requires a narrow beam in the slice, or elevation, plane.

Poor slice resolution leads to partial voluming, which blurs the edges of structures such as vessels and reduces

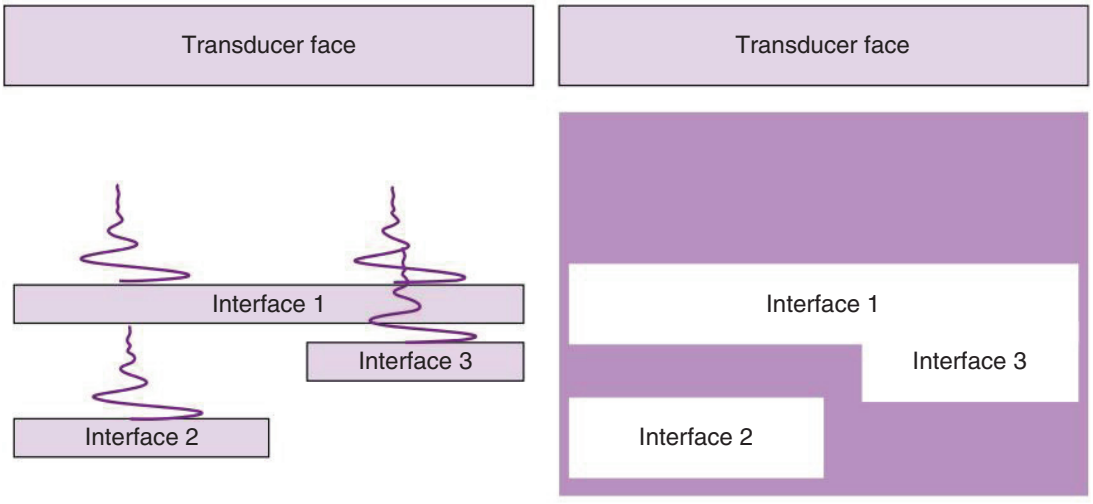

Fig. 7.25 Axial resolution. Interfaces 1 and 2 are far enough apart for the returning echoes not to overlap and allow them to be distinguished in the resulting image. Interfaces 1 and 3 are too close to each other to be distinguished separately and are merged in the resulting image.

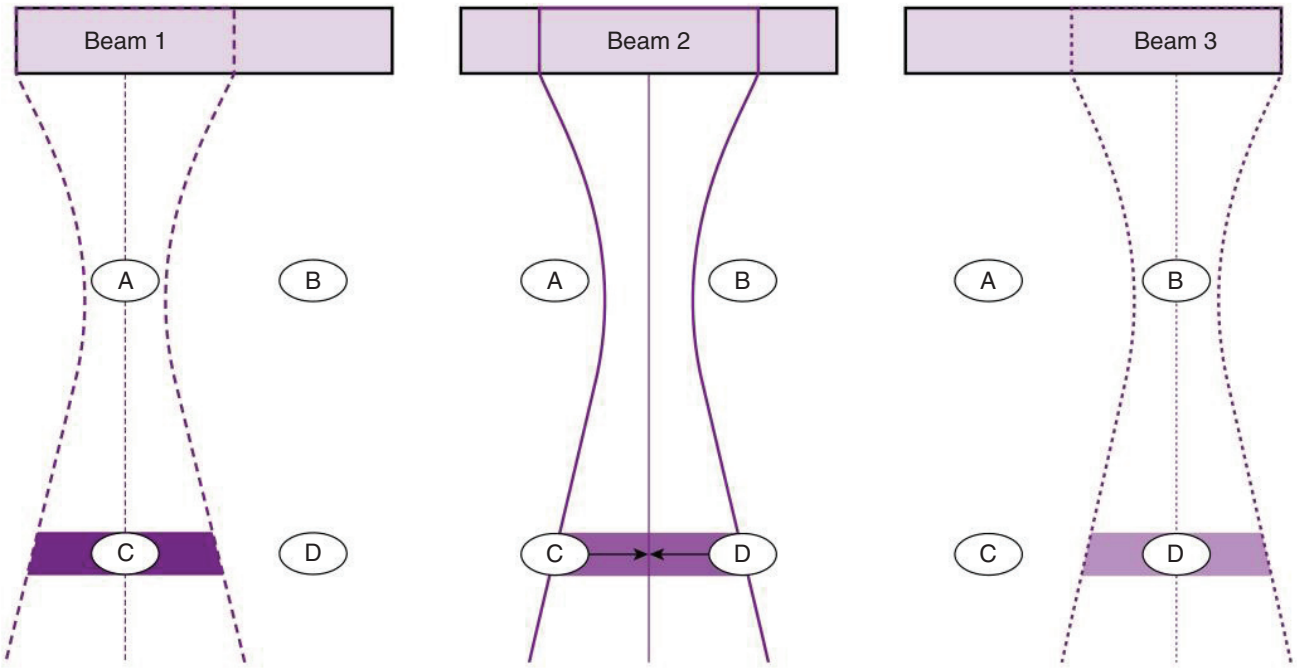

Fig. 7.26 Lateral resolution. Objects A and B are resolvable in the narrow region of beams 1 and 3, but objects C and D are not resolvable in the wider region of beam 2, with C and D merging into a single object.

visibility of small lesions. This is improved with physical focussing, or the additional electronic focussing made possible in 1.5D/3D transducers.

Resolution and Penetration

In choosing a transducer frequency for a particular investigation, it is necessary to compromise between the conflicting requirements of penetration depth (which decreases) and image resolution (which improves) as frequency is increased. Typical figures are:

- 1–8 MHz for general purpose abdominal and cardiac scanning including liver, uterus, and heart
- 5–18 MHz for thyroid, carotid, breast, testis, and other superficial tissues, and for infants
- 10–15 MHz for the eye, which is small and acoustically transparent
- Higher frequencies still may be used in imaging the skin, vessel walls, or joints

IMAGE PROCESSING

Once the raw data is acquired, the image data is processed further to optimise the image for viewing.

The smallest signal that can be detected is just greater than the noise, principally electronic noise. The detected signal dynamic range is the ratio of the maximum to the minimum signal intensity that can be detected. This is typically 70–80 dB, or 40–50 dB after TGC.

The eye can only detect around 30 grey levels, so the actual displayed dynamic range is much smaller, typically around 20–30 dB, displaying around 128–512 grey levels. This is achieved by compression: the data is divided into smaller ranges, each of which is mapped to a reduced range of grey levels (Fig. 7.27).

The amount of compression applied can be chosen depending on the requirements for the resulting image (Fig. 7.28):

- A large displayed dynamic range maps the signals to a large number of grey levels and produces a more flat, low-contrast image, in which subtle changes can be seen.
- A small displayed dynamic range maps the signals to a small number of grey levels and produces a very high-contrast image, in which high-contrast interfaces are enhanced.

The way the input signals are allocated output grey levels can be varied by altering the grey map, which can additionally enhance low-level, medium-level, or high-level signals as required. Reject control is used to filter out low-amplitude noise and scatter.

Temporal averaging (persistence):

For each point in the image, the echoes from 5–10 successive frames can be stored and combined, producing a

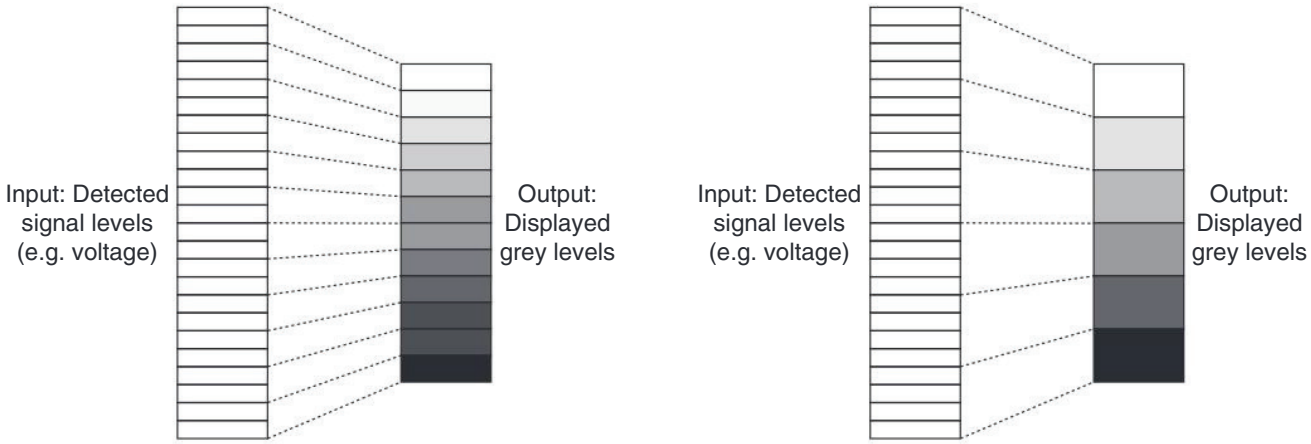

Fig. 7.27 Example of data compression, resulting in reduced dynamic range. Here, the signal range of 24 levels (e.g., voltages) are mapped to the reduced range of 12 grey levels (left) and an even lower dynamic range of 6 grey levels (right).

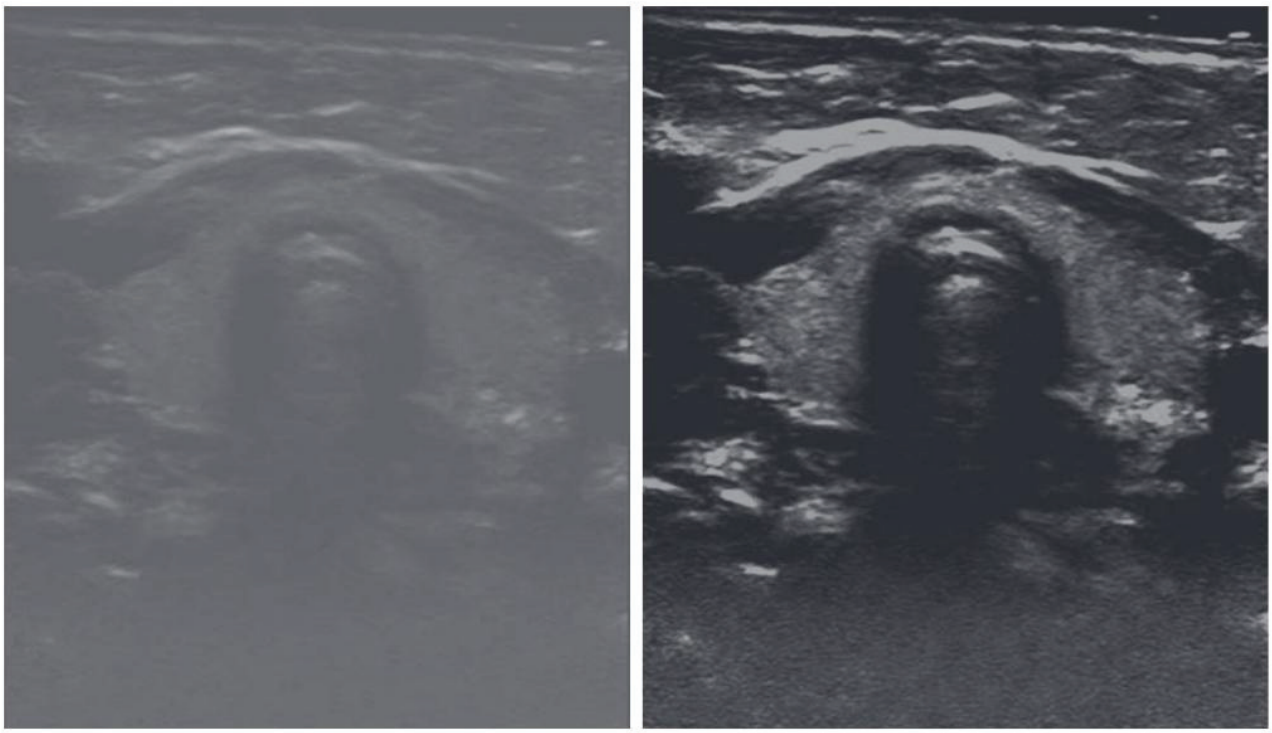

Fig. 7.28 Thyroid images showing high and low dynamic ranges on the left and right, respectively.

time-average value. Because true signals remain constant over time, these are enhanced and randomly varying noise is suppressed, producing a smoother image with increased SNR.

Spatial averaging (spatial compounding):

In this type of averaging, multiple images are produced with the beams steered over a range of angles and combined. Because true signals remain constant over time, these are enhanced and randomly varying noise is suppressed, producing a smoother image with increased SNR.

Image enhancement:

Once the image is acquired, post-processing can be applied to either enhance or suppress certain features that aid or impede image interpretation. This can achieve effects such as speckle reduction or edge enhancement. This often takes the form of a **spatial filter**, which reassigns pixel values

based on the values of surrounding pixels within the image. A smoothing filter, such as an averaging filter, assigns a pixel a value that is similar to that of the surrounding pixels, blurring image features such as speckle. A sharpening filter does the opposite, enhancing edges by increasing the weighting of pixels that contribute to those edges.

ARTEFACTS

Image formation assumes that the ultrasound:

- travels in straight lines
- speed is constant.
- attenuation is constant.
- is reflected only once from each interface.
- is transmitted in single beams perpendicular to the transducer face.

None of these assumptions hold exactly, leading to the presence of artefacts. These artefacts can lead to misdiagnosis or sometimes, when recognised, can aid in diagnosis.

Speckle:

This is produced when small structures within tissue parenchyma are too small and close together to be resolved. Instead, they scatter the ultrasound, and the scattered waves interfere, producing a textured pattern. This pattern is random and unrelated to the actual tissue texture but may be sufficiently characteristic to assist in tissue differentiation.

Reverberation:

Pairs of strongly reflecting interfaces (including transducer and skin or transducer and air — Fig. 7.29) enable multiple reflections back and forth between them before they are fully attenuated. These produce a series of delayed echoes, equally spaced, that falsely appear to be more distant structures. Comet-tail artefacts occur when these reflectors are very close, causing a tapering 'tail' of reflections distal to the reflectors (e.g., in the presence of calcifications).

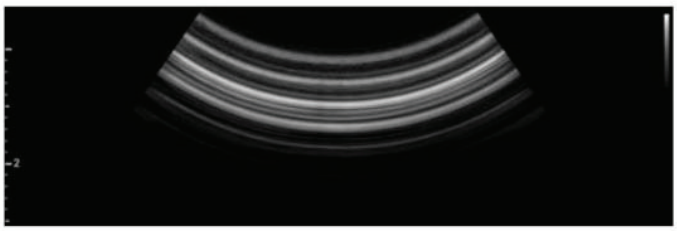

Fig. 7.29 Example of a curvilinear probe operating in air, showing multiple bands due to reverberation back and forth within the probe face.

Ring-down:

When a small gas bubble resonates, it emits ultrasound continuously, resulting in a bright track of signal throughout the scan (e.g., air in the stomach).

Mirroring:

Large, strongly reflective interfaces can act like a mirror, making structures appear displaced onto the opposite side of the interface (e.g., the diaphragm can reflect structures from the liver into the lung).

Acoustic shadowing/enhancement:

Strongly attenuating or reflecting structures reduce the intensity of distal echoes and appear to cast shadows (acoustic shadowing), e.g., bowel gas (Fig. 7.30), lung, bone, gallstones, or kidney stones. Weakly attenuating structures reduce the intensity of distal echoes less than surrounding structures and appear to produce a 'negative shadow', known as posterior acoustic enhancement (e.g., fluid-filled structures such as cysts or a filled bladder; Fig. 7.30). The scanner may try to compensate for these effects using TGC, which can be detrimental to the rest of the image, so this may need to be adjusted in the presence of acoustic shadowing or enhancement.

Distortion:

Refraction of a beam falling obliquely on an interface between tissues with very different speeds of sound displaces the beam and the images of structures beyond, causing distortion (e.g., between two surfaces of bone such as skull). Where the speed of sound is different from the assumed speed, some structures will appear shorter (e.g., lung: slower) or larger (e.g., gallstones: faster). Some scanners have variable speed of sound, such as those used in breast tissue where the sound travels slower.

Side lobes and grating lobes:

Additional, lower-intensity beams can be created which project at an angle from the central beam. These are either in the form of:

- Side lobes: caused by vibrations at the edge of the transducer element
- Grating lobes: interactions between the beams from adjacent transducer elements

If there are strong reflectors in these beams, their reflections are assumed to originate from the central beam, producing ghost structures that are displaced from their correct position.

Harmonic imaging:

As ultrasound propagates through tissue, the speed of sound varies through different tissues, so some parts of the

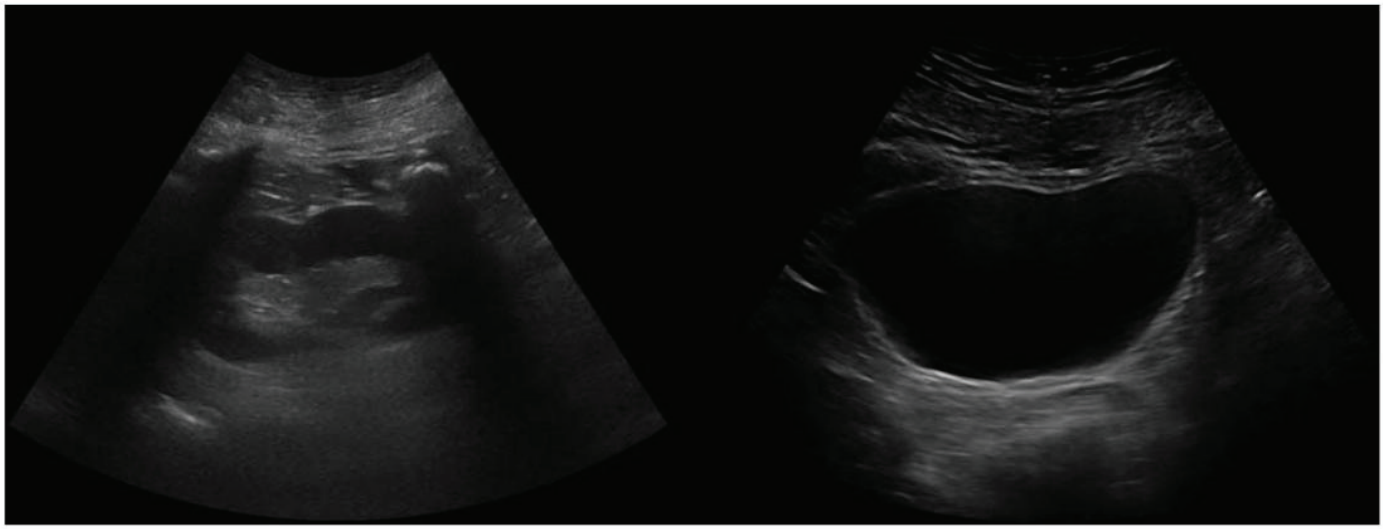

Fig. 7.30 Image of kidney showing acoustic shadowing due to bowel gas (left) and image of bladder showing postcystic enhancement (right).

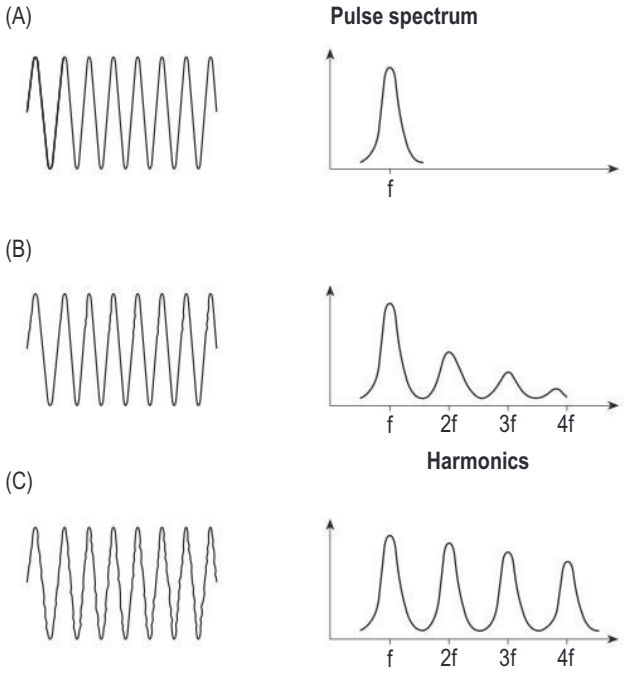

Fig. 7.31 Change in sound wave profile with different tissues at depth: (A) initial, (B) after a few centimetres of tissue, and (C) after a few more centimetres of tissue.

wave 'get ahead' or 'fall behind', distorting the wave. Hence the frequency components also change, containing not only the fundamental frequency, (first harmonic: *f*), but also multiples of this frequency (higher-order harmonics) (Fig. 7.31 For example: for a probe operating at 2 MHz, the 2nd, 3rd, and 4th harmonics (2*f*, 3*f*, and 4*f*) would be 4, 6, and 8 MHz, respectively. This effect becomes more pronounced with depth as the wave passes through more tissue.).

In harmonic imaging, the fundamental frequency (1st harmonic) is transmitted, and both the fundamental and higher-order harmonics are reflected back to the transducer. The 2nd harmonic can produce ultrasound of a sufficient magnitude to be useful. This is separated from the fundamental to produce an image of the harmonic only. There are a variety of methods to separate the 2nd harmonic.

Band filtering: In band filtering (Fig. 7.32A), the fundamental frequency is removed using a low-pass filter. To use this technique, both the fundamental and harmonic frequencies must be contained within the probe bandwidth, but to be separable they cannot overlap, so a longer pulse is needed to produce a narrower transmission band. A narrower bandwidth requires longer pulses, so axial resolution is degraded (although this is compensated for in part by the 2nd harmonic).

Pulse inversion: In pulse inversion (Fig. 7.32B), two pulses are transmitted: one 180° out of phase from the other. On reception, the fundamental components of the signal destructively interfere, and the 1st harmonic components constructively interfere (doubling the received amplitude), enabling imaging with the harmonic only.

This allows axial resolution to be maintained, but doubles the number of pulses required, so reducing the frame rate.

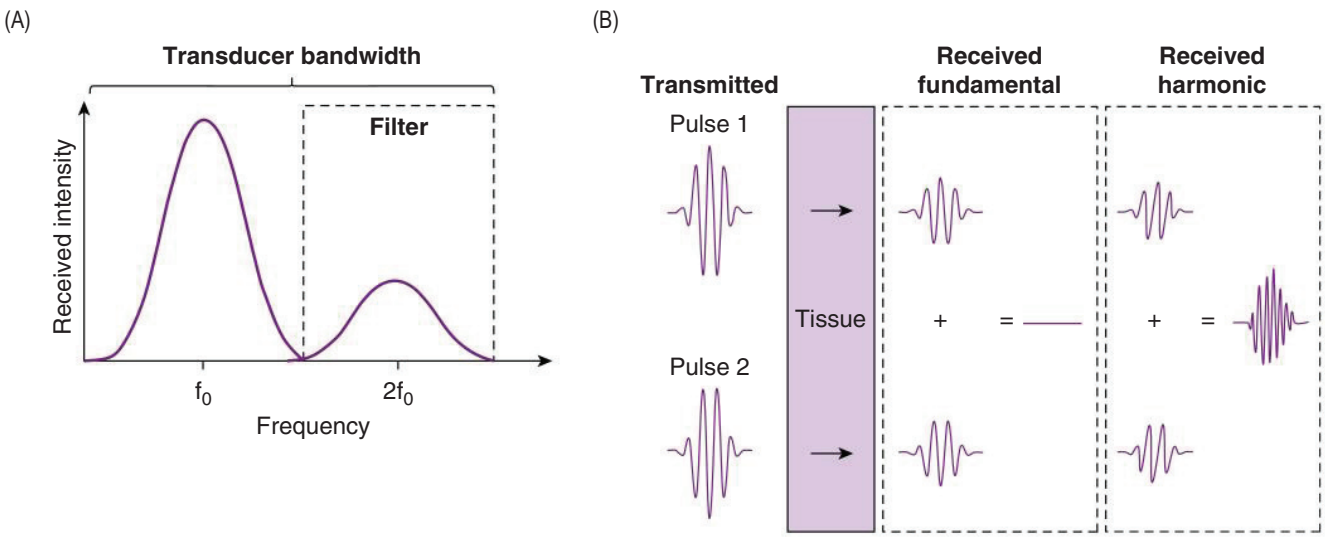

Fig. 7.32 Harmonic imaging techniques: (A) band filtering, (B) pulse inversion.

Other techniques can be used to remove the fundamental signal, including differential harmonics and amplitude modulation.

Advantages of tissue harmonic imaging include the following:

- The amplitude of the harmonic is proportional to the square of the fundamental, so small differences between signal amplitudes are enhanced. This leads to reduction or elimination of low-amplitude artefactual signals or 'clutter' such as those produced by reverberation or sidelobes.
- Different tissues produce different levels of harmonics, enhancing contrast resolution, particularly between tissue and fluid-filled cavities or cysts.

CONTRAST AGENTS

As in computed tomography and magnetic resonance imaging, contrast agents can be used in ultrasound to improve image quality and add information.

The main contrast agents used in ultrasound contain tiny, encapsulated bubbles of gas, or *microbubbles*: either air or low-solubility gas bubbles encapsulated in albumin or liquid shells. These have very low toxicity and are usually destroyed within a few hours and readily eliminated by the body, so are considered generally safe.

The diameter of microbubbles (<4 μm) is much lower than the ultrasound wavelength used but resonates at ultrasound frequencies and their harmonics. This means that microbubbles produce strong signals when they are subjected to ultrasound pulses transmitted at an appropriate frequency. The high levels of harmonics produced allow suppression of surrounding tissues by using harmonic imaging.

Commercially available contrast agents are usually produced from a solution or mixture that is agitated to produce the microbubbles and injected intravenously to examine blood flow or perfusion.

They are destroyed by higher-intensity ultrasound, so they must be imaged at lower powers. A short burst of high-intensity ultrasound can be used to destroy all the microbubbles in the field of view to study refill dynamics.

Applications include:

- Characterisation of lesion vascularity in the liver and kidneys by examining the changes in enhancement over time.
- Increasing the visibility of blood in the heart to enhance the appearance of endocardial borders.
- Assessment of peripheral vascular disease.

Other types of ultrasound contrast agent include perfluorocarbon nanoparticles for imaging of metastases or gold-bound colloidal microtubes that may be immunologically targeted. Targeted contrast agents have the potential not only for diagnosis, but also for targeted drug delivery: microbubbles are injected and then burst using a pulse of ultrasound at the target site.

DOPPLER

The Doppler effect is present whenever a sound or light wave is moving relative to the observer. When incident sound waves I of frequency f are reflected at right angles by

an interface, the wavelength is altered (with speed remaining constant), depending on speed and direction of movement of the interface:

- Movement towards the transducer compresses the wave, hence wavelength decreases and frequency increases.
- Movement away from the transducer dilates the wave, hence wavelength increases and frequency decreases.

The Doppler shift ($f - f'$) is the *change* in frequency and is proportional to the velocity of the interface, v:

$$\frac{(f - f')}{f} = 2 \times \frac{v}{c}$$

hence increasing transducer frequency or speed of interface will increase Doppler shift.

Example: for $v = 30$ cm/s, $f = 10$ MHz, $c = 1540$ m/s

$$(f - f') = 2 \times f \times \frac{v}{c} = 2 \times (10 \times 10^6) \times \frac{30 \times 10^{-1}}{1540}$$
$$= 4 \text{ kHz}$$

Two aspects of the Doppler shift can be measured:

- Magnitude of change in frequency gives velocity of movement
- Measuring whether this is an increase or decrease gives the direction of movement (towards or away from the transducer)

The above refers motion in the direction of the sound only. To account for oblique angles, the term $\cos\theta$ is added (where θ is the angle of insonation — see Fig. 7.33).

$$\frac{(f - f')}{f} = 2 \times \frac{v}{c} \cos\theta$$

The maximum Doppler shift is achieved when $\cos\theta = 1$, i.e., $\theta = 0°$. For movement perpendicular to the transmitted beam, i.e., $\theta = 0°$, $\cos\theta = 0$, producing no Doppler effect.

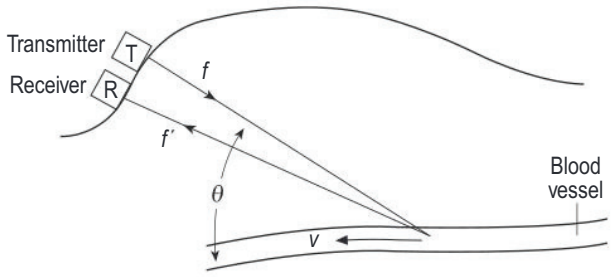

Fig. 7.33 Continuous wave Doppler: detection of blood flow in a vessel.

In diagnostic ultrasound, Doppler is usually used to measure blood flow, by measuring the signal backscattered by blood cells.

The Doppler signal can be displayed on a screen as a spectrum, or output as audio. The Doppler shift frequency is much smaller than the transmitted frequency (within 0—10 kHz) and within the audible range, meaning that the Doppler signal can be amplified and output through a speaker: a higher pitch indicates faster flow, and a harsher sound indicates greater turbulence.

Continuous Wave (CW) Doppler

CW is the simplest form of Doppler (Fig. 7.33). Typically, it uses two slightly angled transducers: one for transmission (T), the other for reception (R). The transmitted frequency is suppressed and the Doppler shift is extracted electronically. This technique is not pulsed, so depth discrimination is not possible. Applications include monitoring the foetal heartbeat.

CW Doppler uses a high Q with no backing block to produce a precise, narrow frequency bandwidth with high output resulting in good accuracy in Doppler shift measurement. Typical frequencies used are within range of 2—10 MHz, depending on the vessel depth.

Pulsed Wave (PW) Doppler

In contrast to CW Doppler, PW Doppler enables measurement of the Doppler shift within a specified volume by using ultrasound pulses that allow depth discrimination, as for B-mode.

This technique is performed using a standard B-mode transducer, with a narrow section used for Doppler, as follows:

1. The B-mode image is produced and used to choose a line of sight for the Doppler beam.
2. Cursors along this line are chosen to define a *sampling volume*, usually positioned over the vessel in which the blood flow is to be measured (Fig. 7.34).
3. The angle of flow is specified by the user to enable accurate velocity calculation.
4. Doppler pulses are transmitted along this line to produce a Doppler signal.
5. Only echoes arriving within a short interval. The timing of this acceptance window defines the depth and width of the sampling volume, or *gate*.

Example: Ultrasound takes ~7 μs to travel 1 cm in tissue. If the range gate is opened at 70 μs and closed at 77 μs, then, accounting for the round trip:

$$70\mu s \gg \frac{1}{2} \times \frac{70}{7} \times 1 = 5\,\text{cm};\ 77\mu s \gg \frac{1}{2} \times \frac{77}{7} \times 1$$
$$= 5.5\text{cm}$$

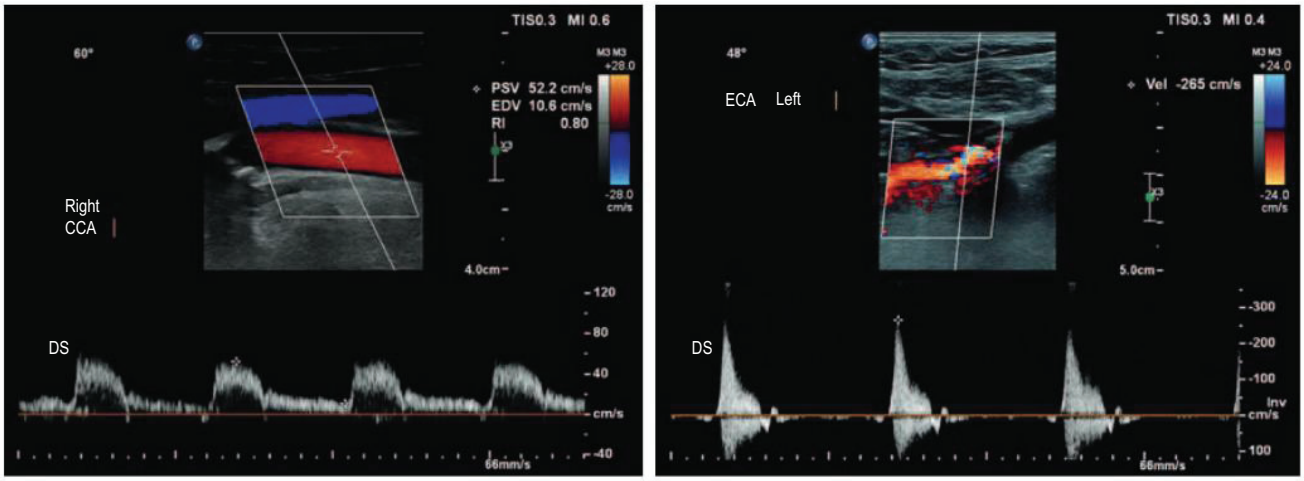

Fig. 7.34 Triplex images with the colour Doppler box superimposed onto the B-mode image, the pulsed wave Doppler range gate selected using the B-mode image, and the Doppler sonogram (DS) displayed below. Common carotid artery (CCA) on the left showing patent flow in both directions, and external carotid artery (ECA) on the right showing turbulent flow and aliasing.

Hence, blood velocities will be sampled in a volume of tissue about 5 mm thick, starting at a depth of 5 cm.

The Doppler signal comprises a wide range of audio frequencies corresponding to the range of blood velocities in the sampling volume. This can be analysed and displayed as a time-velocity spectrum or sonogram (Fig. 7.35). The transmitted Doppler frequency is used to maximise resolution, as lower frequencies allow faster flow to be measured.

Like in B-mode, the PRF depth and depth range are balanced:

- A higher PRF is used for more superficial vessels.
- The PRF is reduced as the range is increased to ensure that successive pulses do not overlap.

Real-Time Colour Flow

Whereas a greyscale (B-mode) image shows the strength of echo coming from each pixel, a *colour-mapped Doppler* image shows the direction and speed of movement or flow occurring in each pixel. The Doppler pulse is longer than that used in B-mode and the time along each line allows fewer pulses per line than PW Doppler (around 4—12 consecutive pulses in colour versus 100 in PW), so that a compromise is made between depth discrimination and accuracy of flow measurement. For this reason:

- The colour Doppler image usually covers a smaller field of view and has lower spatial resolution than the B-mode image.
- The data only allow estimation of the mean and variance (as a measure of turbulence) of velocity in each sample volume and to colour each pixel accordingly.

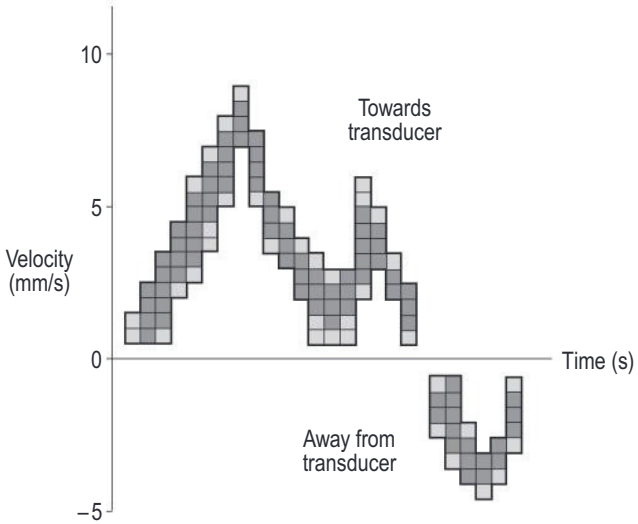

Fig. 7.35 Sonogram, showing range of velocities within the sampling volume.

The colour map is superimposed onto a user-defined region of the greyscale B-mode image using a colour scale to denote direction and speed, for example:

- Flow towards the transducer: red
- Flow away from the transducer: blue
- High-flow variance (often indicating turbulence): yellow or cyan

The performance of colour Doppler is limited by the short time available to collect the data from each beam position. The following factors can be varied and are interrelated, so should be carefully balanced:

- Frame rate, which should be fast enough to follow changes in flow velocity.
- Penetration depth, which is inversely proportional to PRF.
- Field width, or sector which can be reduced to increase frame rate.
- Line density, which should be high enough for good spatial resolution.
- PRF, which can be controlled using the colour scale and should be high enough to give accurate velocity information.

The combinations of two or three imaging methods such as B-mode and PW Doppler and/or colour Doppler are known as *duplex or triplex imaging*, respectively. This is achieved by transmitting bursts of Doppler pulses between B-mode pulses.

Doppler Aliasing

In pulsed Doppler modes, the Doppler shift is measured by detecting the phase of successive pulses. To accurately measure the full range of frequencies in this way, the sampling rate must be high enough to satisfy the Nyquist criterion, i.e., the PRF must satisfy:

$$PRF \geq f_{max}$$

Where f_{max} is the maximum Doppler shift in the sampled volume. High-velocity flow outside of this limit therefore results in aliasing, manifesting as flow in the opposite direction. This appears as follows:

- In PW Doppler, the top of the spectrum wraps around from the top to the bottom of the trace (or vice versa).
- On colour Doppler, flow that should appear red changes to black and then blue. Fast laminar flow appears as an aliased blue centre with a red edge (or vice versa) and high-speed jets with associated turbulence appear as a coloured mosaic.

Aliasing can be resolved by increasing the Doppler scale to increase the PRF, or by switching to CW Doppler, which does not suffer from aliasing.

If a high enough PRF is not possible at the depth of flow examined, this can be partially overcome using high PRF mode. As in Fig. 7.36, instead of only sampling from a single volume A, doubling the PRF samples in both volume A and another at half the depth, B. Increasing the PRF further samples from even more volumes. This method relies on appropriate positioning by the operator to ensure that only one sample volume is positioned over an area of flow.

Power Doppler

Power Doppler images do not give any indication of speed or direction, but instead maps the *amplitude* of the Doppler signal. A higher density of red blood cells produces a greater amplitude signal, which appears brighter on the image. In this case, all movement, regardless of phase, contributes to the amplitude, so the power Doppler image reflects the *quantity* of blood flow.

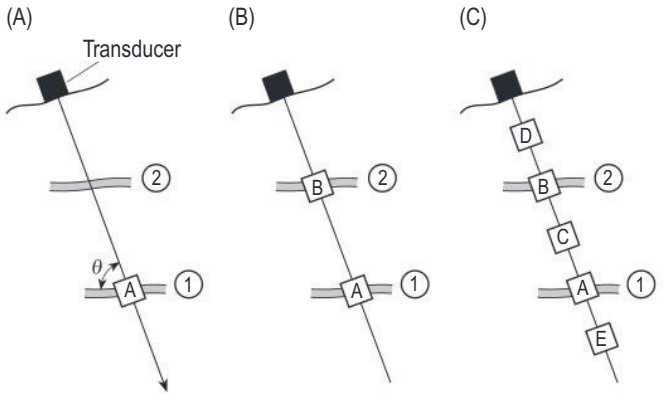

Fig. 7.36 Pulsed Doppler: (A) normal and (B and C) high pulse repetition frequency modes; 1 and 2 are blood vessels.

The main application of power Doppler is to differentiate between areas with and without flow, often where the flow is slow, like in imaging vessel walls. It is also much more sensitive than colour Doppler, enabling imaging of smaller vessels and perfusion of organs such as the kidneys. This technique does not suffer from aliasing as it does not measure velocity, but its high sensitivity does make it susceptible to artefacts caused by the motion of surrounding tissue.

ELASTOGRAPHY

As well as imaging, ultrasound can be used to examine the elasticity or stiffness of tissues. There are two main techniques: strain imaging and shear wave elastography.

Strain imaging: Ultrasound imaging is performed before and during compression of the tissue, using either:

1. Free-hand compression
2. Mechanical compression
3. Internal physiological movement such as cardiovascular or respiratory
4. Acoustic radiation force impulse (ARFI): using a short, high-intensity ultrasound pulse to produce a radiation force that causes tissue displacement

The tissue displacement caused by the applied stress is estimated to determine the deformation (i.e., stiffer tissues deform less than more elastic tissues). This is used to produce a colour map of strain distribution, showing tissue hardness, which is superimposed onto a B-mode image. Because it is not possible to quantify the stress applied in techniques 1–3, these produce qualitative results and cannot be used for absolute measurement of tissue stiffness.

Shear wave elastography: The speed of sound depends on tissue stiffness, therefore by measuring the speed, the stiffness can be quantified. Shear waves travel much more slowly in soft tissue than longitudinal compression waves: approximately 1−10 m/s. This allows the shear wave propagation to be imaged as it propagates (usually with A-mode ultrasound), allowing its speed to be calculated. Ultrasound imaging is performed before and during the generation of shear waves in the tissue, using either:

1. Mechanical vibration of the probe
2. ARFI

The calculated shear wave speed is converted to a quantitative measure of tissue stiffness. Depending on the technique employed, this can be done using A-mode or B-mode, to produce a quantitative measurement for a chosen area, or two-dimensional colour map.

SAFETY CONSIDERATIONS

Ultrasound is not an electromagnetic radiation and is non-ionising. Used in diagnosis, it is a low-risk and low-cost method of medical imaging. However, harmful biological effects have been identified at exposure levels more usually associated with ultrasound therapy.

There has been no confirmed evidence of damage from diagnostic ultrasound exposure. However, the output of each probe should be checked periodically, and operators should keep within safety guidelines (see below), to minimise the potential for effects such as the following:

Thermal effects:

Local heating due to frictional, viscous, and molecular relaxation processes, leading to chemical damage but mitigated by blood flow. The risk from this effect is increased for patients who already have a raised body temperature, or in more sensitive tissues including the embryo; the head, brain, or spine of a foetus or neonate; and the eye.

Mechanical/Nonthermal Effects

- **Cavitation:** The high peak pressure changes can cause microbubbles in a liquid or near liquid medium to oscillate (expand and contract). If the expansion results in very sudden collapse (inertial cavitation), this can cause cellular damage. This is more likely at high pressures and low frequencies and if pulses are long enough to allow resonance to be reached. It is also more likely in the presence of existing gas such as in the lung or bowel or when using microbubble contrast agents.
- Acoustic streaming of cellular contents in the direction of the beam, affecting cell membrane permeability.
- Mechanical damage to cell membranes caused by violent acceleration of particles.

Output Measures and Safety Indices

The intensity of an ultrasound beam is greatest in the focal region. Typically, the time-averaged intensity (I_{SPTA}: the spatial peak averaged over the examination) is around 0.1 W/mm^2. The peak intensity (I_{SPTP}: the intensity at the time during the brief pulse when it is at the peak) is likely to be 1000 times greater. The I_{SPTA} will increase for longer pulses, such as those used in Doppler.

Thermal effects are estimated using the thermal index (TI), which gives an indication of the temperature rise in tissue:

$$TI = \frac{\textit{Emitted Power}}{\text{Power required to increase the temperature by } 1^{\circ}\text{C}}$$

As the TI increases above the value of 1, exposure times should be reduced appropriately, depending on the application and the condition of the patient. Recommended exposure times are given by the British Medical Ultrasound Society,[1] along with a summary of ultrasound safety and techniques for exposure reduction. For example, scanning at a TI of 2.0 can be performed for up to 60 min, whereas at a TI of 3.0, times should be restricted to 4 min. Further care and more restrictive scan times are needed in obstetric or ophthalmic imaging or when the patient is feverish.

Mechanical effects are estimated using the mechanical index (MI): the maximum amplitude of the pressure pulse, defined as the peak rarefaction (negative) pressure divided by the square root of the ultrasound frequency. As with TI, exam times should be reduced appropriately once MI reaches the threshold of 0.7, at which mechanical effects are thought to occur.

Diagnostic ultrasound equipment output should not exceed the following:

- Derated $I_{SPTA} \leq 720$ mW/cm^2 (this is further reduced for applications such as cardiac, foetal, or ophthalmic imaging)
- MI ≤ 1.9 or derated $I_{SPTA} \leq 190$ W/kg

The calculated indices, MI and TI, which are worst-case-based, should be indicated on the scanner display when the equipment is capable of exceeding an index value of 1, to enable users to continuously monitor exposure and reduce scan times accordingly.

QUALITY ASSURANCE

Ultrasound scanners are generally reliable and stable; however, slight damage to a probe or drift in the electronics can lead to deterioration of the image. Image quality tests to detect faults include the following:

- **Sensitivity:** tested by imaging in air to produce a reverberation image. A reduction in the depth of the last reverberation is an indication of a reduction in sensitivity.
- **Low contrast penetration and uniformity:** tested by imaging a uniform tissue-mimicking material (TMM). Penetration is determined by measuring the depth at which the speckle produced from the TMM is no longer visible. Examining the uniformity of the image can detect other faults such as a broken transducer element.
- **Spatial resolution:** tested by imaging narrow threads or pin targets mounted on a frame and immersed in a fluid in which the speed of sound is 1540 m/s, and measuring the spread axially and laterally of the cross-section. Imaging the spread of these pins allows a rough visualisation of the variation in in-plane beam width with depth and rotating the probe surface by 45°.
- Other image quality tests such as greyscale performance and Doppler function require more complex test objects, often in a TMM containing test objects of varying dimensions and acoustic properties (Fig. 7.37).

The power output of the transducer is measured by 'weighing' the sound pressure with a force balance or by measuring the heating effect using a calorimeter. More sophisticated techniques can be used to measure the intensity or pressure distribution throughout the ultrasound field.

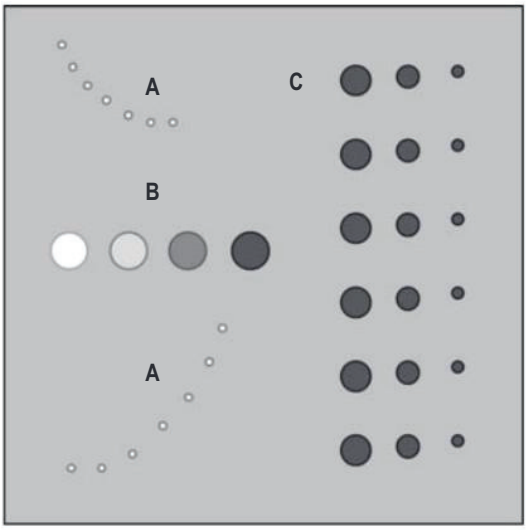

Fig. 7.37 Example tissue-mimicking test object containing targets for measuring (A) resolution, (B) greyscale performance, and (C) anechoic target detection.

Recommendations for routine quality assurance are given by the Institute of Physics and Engineering in Medicine in Report 102.[2]

REFERENCES

1. British Medical Ultrasound Society. *Guidelines for the Safe Use of Diagnostic Ultrasound Equipment*. London, UK: BMUS; 2009.
2. Institute of Physics and Engineering in Medicine. *Report 102: Quality Assurance of Ultrasound Imaging Systems*. York, UK: IPEM; 2010.