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In this chapter the physics of medical ultrasound will be discussed at an introductory level for users of the technology. Several texts exist which readers can consult to deepen their understanding.<sup>1-5</sup> Discussed later in this chapter are parameters related to ultrasound in tissue such as speed of ultrasound, attenuation and acoustic impedance. Values of these parameters for commonly encountered tissues and materials are quoted (for more values see Duck<sup>6</sup> and Hill et al.<sup>7</sup>). The clinical user is not required to have a detailed knowledge of these values but some knowledge helps in the production and interpretation of ultrasound images and Doppler blood flow measurements. Basic physics is also of central importance in considerations of safety.

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## PRODUCTION OF ULTRASOUND

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Ultrasound vibrations (or waves) are produced by a very small but rapid push-pull action of a probe (transducer) held against a material (medium) such as tissue. Virtually all types of vibration are referred to as acoustic, whereas those of too high a pitch for the human ear to detect are also called ultrasonic. Vibrations at rates of less than about 20 000 push-pull cycles/s are audible sound, above this the term ultrasonic is employed. In medical ultrasound, vibrations in the range 20 000 to 50 000 000 cycles/s are used. The term frequency is employed rather than rate of vibration and the unit

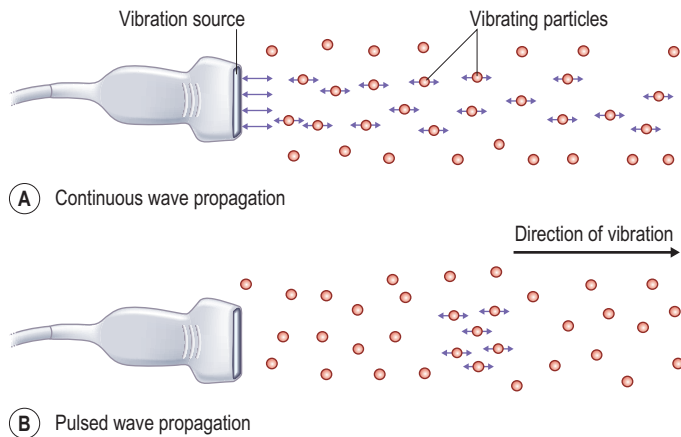
hertz (Hz) rather than cycles/s. We therefore use frequencies in the range 20 kilohertz (20 kHz) to 50 megahertz (50 MHz).

We most commonly encounter audible acoustic waves produced by the action of a vibrating source on air (vocal cords, loudspeaker, musical instruments, machinery). In medical ultrasound the source is a piezoelectric crystal, or several, mounted in a hand-held case and driven to vibrate by an applied fluctuating voltage. Conversely, when ultrasound waves strike a piezoelectric crystal causing it to vibrate, electrical voltages are generated across the crystal, hence the ultrasound is said to be detected. The hand-held devices containing piezoelectric crystals and probably some electronics are called transducers since they convert electrical to mechanical energy and vice versa. They are fragile and expensive, about the same price as a motor car. Transducers are discussed more fully in Chapter 2. The great majority of medical ultrasound machines generate short bursts or pulses of vibration, e.g. each three or four cycles in duration; a few basic fetal heart or blood flow Doppler units transmit continuously. Figure 1.1 illustrates the generation of pulsed and continuous ultrasound. For a continuous wave an alternating (oscillating) voltage is applied continuously whereas for a pulsed wave it is applied for a short time. The basic data for most ultrasound techniques is obtained by detecting the echoes which are generated by reflection or scattering of the transmitted ultrasound at changes in tissue structure within the body.

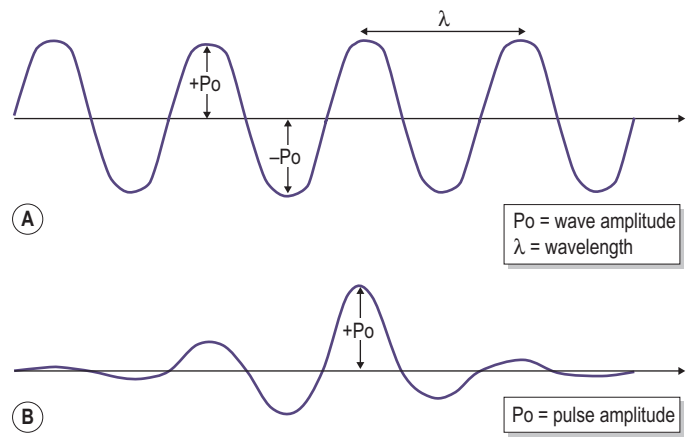
The push-pull action of the transducer causes regions of compression and rarefaction to pass out from the transducer face into the tissue. These regions have increased or decreased tissue density. A waveform can be drawn to represent these regions of increased and decreased pressure and we say that the transducer has generated an ultrasound wave (Fig. 1.2). The distance between equivalent points on the waveform is called the wavelength and the maximum pressure fluctuation is the wave amplitude (Fig. 1.3). If ultrasound is generated by a transducer with a flat face, regions of equal compression or rarefaction will lie in planes as the vibration passes through the medium. Plane waves or wavefronts are said to have been generated. Similarly if the transducer face is convex or concave the wavefront will be convex or concave. The latter can be used to provide a focused region at a specified distance from the transducer face. In tissue if we could look closely at a particular point, we would see that the tissue is oscillating rapidly back and forward about its rest position. As noted above, the number of oscillations per second is the frequency of the wave. The speed with which the wave passes through the tissue is very high close to 1540 m/s for most soft tissue, i.e. several times the speed of most passenger jet planes. We will see in the discussion of imaging techniques that this is very important since it means that pulses can be transmitted and echoes collected very rapidly, enabling images to be built up in a fraction of a second. The speed of sound,  $c$ , is simply related to the frequency,  $f$ , and the wavelength,  $\lambda$ , by the formula:

$$c = f\lambda$$

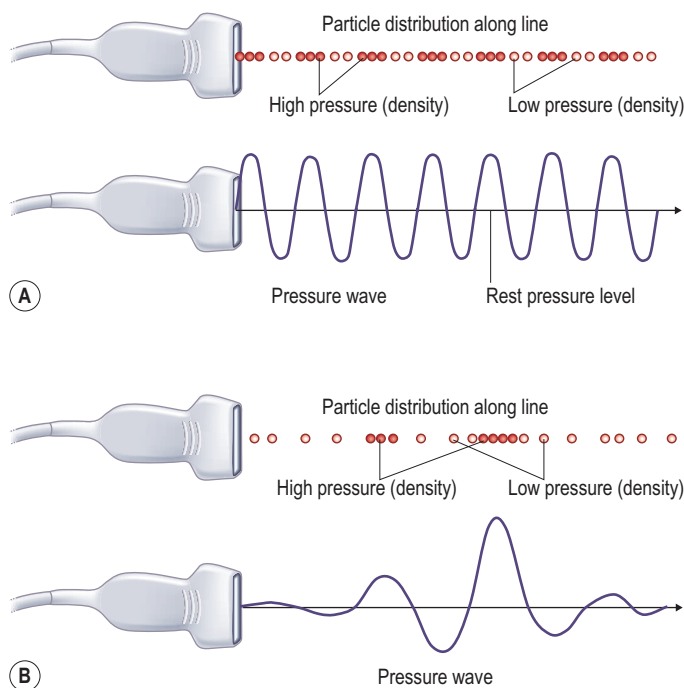
In the acoustic waves just described the oscillations of the particles of the medium are in the same direction as the wave travel. This type of wave is called a longitudinal wave or compressional wave since it gives rise to regions of increased and decreased



**Figure 1.1** The generation of continuous and pulsed wave ultrasound by a vibrating source in contact with the propagating medium.



**Figure 1.3** Ultrasound wavelength and wave amplitude. **A:** Continuous wave. **B:** Pulsed wave.



**Figure 1.2** The waveform description of ultrasound pressure fluctuations. **A:** Continuous wave. **B:** Pulsed wave.

pressure. In another type of wave the oscillations are perpendicular to the direction of wave travel, like ripples on a pond, and they are called transverse or shear waves. At MHz frequencies the latter are attenuated rapidly in tissues and fluids and hence are not encountered at present in diagnostic ultrasound. Ongoing developments at lower frequencies aim to utilise them in elastography (see later in this chapter). Transmitted energy, power and pressure amplitude are discussed more quantitatively in Chapter 4 on safety.

The speed of ultrasound in soft tissue depends on its rigidity and density (Table 1.1), the more rigid a material, the higher the speed. From the table we can see that the speed in soft tissues is fairly closely clustered around an average value of 1540 m/s. Virtually all diagnostic instruments measure the time of echo return after the instant of pulsed ultrasound transmission and then use the speed in tissue to convert this time into the tissue depth of the reflecting structure. For a mixture of soft tissues along the pulse path, an accurate measure of depth is obtained by the assumption of an

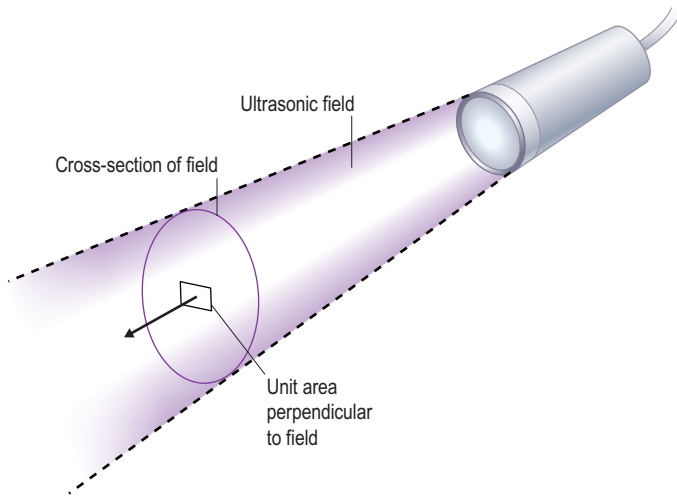
**Table 1.1** Speed of ultrasound and acoustic impedance

Material	Speed (m/s)	Acoustic impedance (g/cm <sup>2</sup> s)
Water (20°C)	1480	1.48 × 10 <sup>5</sup>
Blood	1570	1.61 × 10 <sup>5</sup>
Bone	3500	7.80 × 10 <sup>5</sup>
Fat	1450	1.38 × 10 <sup>5</sup>
Liver	1550	1.65 × 10 <sup>5</sup>
Muscle	1580	1.70 × 10 <sup>5</sup>
Polythene	2000	1.84 × 10 <sup>5</sup>
Air	330	0.0004 × 10 <sup>5</sup>
Soft tissue (average)	1540	1.63 × 10 <sup>5</sup>

Basic ultrasound concepts

- Ultrasound is high-frequency vibration that travels through tissue at high speed, close to 1540 m/s.
- In medical ultrasound, the frequency range used is 1 to 50 MHz.
- Ultrasound vibration travels as a pressure waveform.
- Ultrasound can be generated and detected by piezoelectric crystals contained in small hand-held transducers.
- Echoes produced by transmitted ultrasound pulses at tissue interfaces are the basic source of information in diagnostic ultrasound.

average speed of 1540 m/s for conversion of time into depth. The high speed in bone can cause severe problems as will be seen later when refraction of the ultrasound is considered. Unless bone is thin it is best avoided during ultrasound examinations. The speed of ultrasound in soft tissue is independent of frequency over the diagnostic range, i.e. 1 to 50 MHz. The very high value of the speed of sound in tissue means that echo data can be collected very rapidly; e.g. for a particular beam direction echoes might be collected in less than 0.2 millisecond. This rapid collection could allow a single image (frame) to be produced in 20 milliseconds. In a later section it will be seen that the speed of sound in blood is also used in the Doppler equation when the velocity of blood is calculated.



**Figure 1.4 Intensity at a point** – rate of flow of energy through unit area in an ultrasound beam. Power – rate of flow of energy through whole cross-section of ultrasound beam.

## INTENSITY AND POWER

When a transducer is excited by an electrical voltage, vibrations pass into the tissue, i.e. energy passes from the transducer to the tissue. The intensity at a point in the tissue is the rate of flow of energy through unit area at that point (Fig. 1.4). For example, the intensity may be 100 milliwatts per square centimetre (100 mW/cm<sup>2</sup>). The intensity is related to the square of the pressure wave amplitude.

The region of tissue in front of the transducer subjected to the vibrations is referred to as the ultrasound field or beam. Intensity is often measured at the focus of the field or within 1 or 2 cm of the transducer face. Transducer designers will often measure the pressure amplitude or intensity over the full depth range for which the transducer will be used.

For safety studies, intensity is often defined precisely to ensure validity in the conclusions drawn; for example,  $I_{\text{spk}}$  is the intensity at the spatial peak (often the focus) averaged over time. Each definition gives rise to an intensity relevant to that definition. These intensities are not used in routine scanning but it is useful to know their origins when the safety literature is being studied (see Appendix at the end of this chapter). It will be seen in Chapter 4 that from a clinical application point of view, the two quantities of interest to optimise safety are thermal index (TI) and mechanical index (MI).

The power of an ultrasonic beam is the rate of flow of energy through the cross-sectional area of the beam (Fig. 1.4). This is a quantity that is often quoted since it gives a feeling for the total output and is relatively easy to measure.

Power and intensity controls on ultrasonic scanners are sometimes labelled using the decibel notation. This occurs less frequently on new machines. Although not essential for the operation of machines, familiarity with this labelling removes its mystique and can be helpful when reading the literature. The manipulation of machine controls is covered in Chapter 3, where it is demonstrated that logical operation can be readily achieved. The decibel notation basically relates the current value of intensity or power to a reference value. For example, taking the maximum intensity position of the control as giving the reference output value  $I_0$ , other values of intensity  $I$ , corresponding to other control positions, are calibrated relative to  $I_0$  as:

$$\text{Output intensity in decibels (dB)} = 10 \log(I/I_0)$$

This output in dB is therefore the ratio of two intensities and is not an absolute unit like cm or watt. Output power controls can also be labelled in terms of dB:

### Ultrasound beams and power

- The vibration produced by the transducer is a flow of energy through the tissue.
- Intensity and power are related but different physical quantities that are measures of the flow of energy and are of particular interest with regard to safety.
- The ultrasound field or beam is the region in front of the transducer that is affected by the transmitted vibration.
- The decibel notation can be employed to label the controls of intensity, power and amplifier gain. Decibel labelling of a control is not absolute but refers each setting of the control to a reference level.
- Radiation force is experienced when the transmitted energy strikes a target and is reflected or absorbed.

$$\text{Output power in decibels (dB)} = 10 \log(P/P_0)$$

The dB notation was employed in the past when it was fairly difficult to measure absolute values of intensity and power in units of mW/cm<sup>2</sup> or mW respectively and also since separate calibration is required for each transducer. This notation is essentially an historical hangover and is of no value in machines designed for clinical application. More recently the outputs of machines have been related to possible biological effects, in particular heating and cavitation. Cavitation is the violent response of bubbles when subjected to the pressure fluctuations of an ultrasound wave (Chapter 4). Thermal (TI) and mechanical indices (MI) relate to these phenomena and are displayed onscreen (Chapter 4).

The decibel notation is also found applied to the gain of a receiver amplifier:

$$\text{Gain} = \text{Output volts}/\text{Input volts}$$

Or:

$$\text{Gain in decibels} = 20 \log(V_{\text{out}}/V_{\text{in}})$$

Another application relates to differences in pressure amplitude of an ultrasound wave:

$$\text{Acoustic pressure difference in decibels} = 20 \log(p_1/p_2)$$

Absolute values of ultrasonic intensity are normally measured with a hydrophone, which takes the form of a small probe with a piezoelectric element at its tip.

Absolute values of ultrasonic power are measured with a radiation power balance which is based on the following. There is a flow of energy in an ultrasonic field from the transducer as the vibrations pass into water. Associated with the flow of energy in a beam there is a flow of momentum. Due to this momentum a force is experienced by an object placed in the field, the force being directly related to the power of the field. These small forces can be measured by placing the object under water on the pan of a sensitive chemical balance, hence the name power balance.

For a beam of power  $P$  incident perpendicularly on a totally reflecting flat surface:

$$\text{Radiation force} = 2P/c$$

The force experienced is 0.135 mg per milliwatt. Power levels down to 0.1 mW can be measured with balances. For an object that totally absorbs the beam, the force is half that of the reflection case.

## DIFFRACTION AND INTERFERENCE

Diffraction describes the spreading out of a wave as it passes from its source through a medium. The pattern of spread is highly dependent on the shape and size of the source relative to the

wavelength of the sound. To produce an approximately parallel-sided beam, the diameter of the crystal face is typically 10 to 20 times that of the ultrasound wavelength. The diffraction pattern of a disc-shaped crystal as found in basic medical ultrasound transducers is approximately cylindrical for a short distance, after which it diverges at a small angle (Fig. 1.5). The diffraction pattern from a small transducer is divergent at a larger angle from close to the transducer (Fig. 1.6A). Within a diffraction pattern there may be fluctuations in intensity, particularly close to the transducer. Diffraction also occurs beyond an obstacle such as a slit aperture or an array of slits which partially blocks the wavefront (Fig. 1.6B). The narrowness of a beam or the sharpness of a focus is ultimately determined by diffraction. The submillimetre wavelengths associated with the high frequencies of medical ultrasound permit the generation of well-focused beams, one of the most important facts in medical ultrasound technology. The higher the frequency, the narrower the beam can be made and hence the finer the image detail. Unfortunately, we will see later that absorption also increases with frequency, which puts an upper limit on the frequency that can be employed in any particular application.

Interference of waves occurs when two or more overlap as they pass through the propagating medium. The resultant wave pressure amplitude at any point is determined by adding the pressure amplitudes from each wave at the point (Fig. 1.7A). When the waveforms are in step, they add constructively (constructive

interference) to give an increase in amplitude. Out of step they add destructively (destructive interference), resulting in a lower pressure amplitude (Fig. 1.7B). This ‘principle of superposition’ is widely used to predict ultrasound field shapes using mathematical models. In such modelling the shape of the transducer field is calculated by considering the crystal face to be subdivided into many small elements. The diffraction pattern for each element is calculated and the effect of them overlapping and interfering gives the resultant ultrasound field shape. Interference and diffraction are rarely considered in detail since, as techniques stand at present, ultrasound beams and fields are not tightly defined.

The terms ‘ultrasound field’ and ‘ultrasound beam’ tend to get used interchangeably. It would probably be better if we restricted ‘field’ to label the transmitted ultrasound pattern in front of the transducer and ‘beam’ to a combination of the transmitted field and the reception sensitivity pattern in front of the transducer. For a single crystal transducer the two patterns are the same but for array transducers they are probably always different. For example, the transmission focusing and the reception focusing are usually different (Chapter 2). It is the shape of the beam, i.e. the combination of transmission and reception, that plays an important part in determining the detail in an image. The loose usage of ‘field’ and ‘beam’ is not usually serious, but it is worth remembering that both transmit power and receiver gain influence the lateral resolution in an image. Image detail (resolution) is discussed later in this chapter.

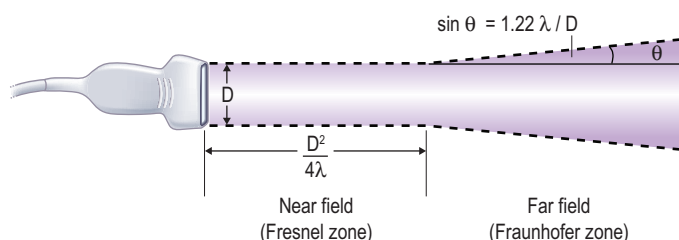


Figure 1.5 Schematic diffraction pattern for a disc-shaped crystal generating a continuous wave beam.

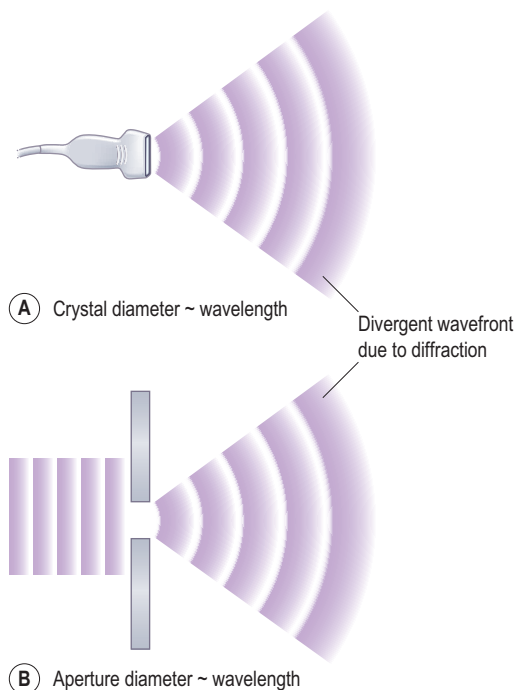


Figure 1.6 Diffraction pattern. A: For a small transducer. B: For a small aperture.

## IMAGE SPECKLE

When an ultrasound pulse passes through tissue the very large number of small discontinuities generate small echoes which travel back to the transducer and are detected. These small echoes overlap and interfere both constructively and destructively to produce a fluctuating resultant signal (Fig. 1.8). The related electronic fluctuating signal produced by the transducer is presented along each scan line in the display as fine fluctuations in the grey shades. The whole image appears as a speckle pattern. In practice the speckle is often a combination of the true speckle from very small tissue structures

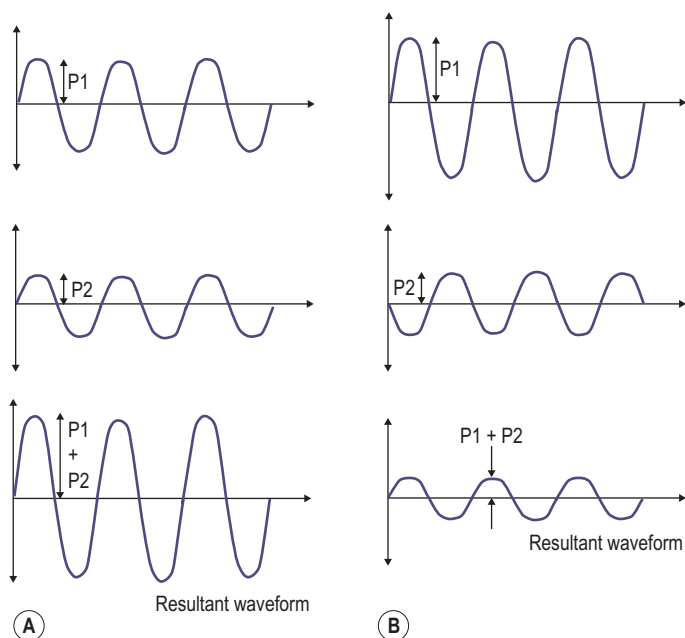
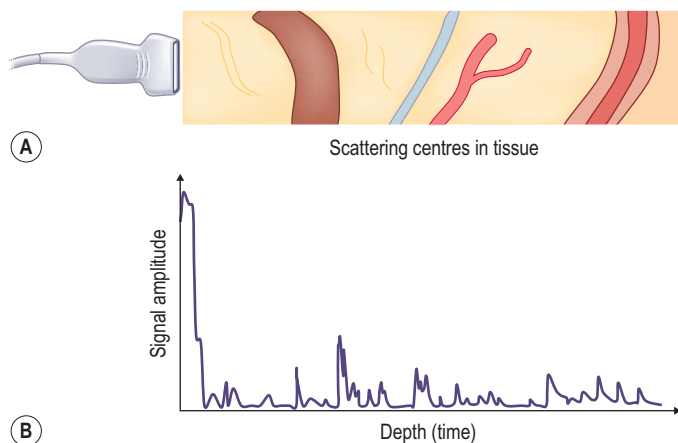


Figure 1.7 A: Constructive interference of two waves. B: Destructive interference of two waves.



**Figure 1.8** Fluctuating echo signals resulting from constructive and destructive interference of echoes from structures lying along one beam direction through tissue. **A:** Transducer transmitting ultrasound along one beam direction into complex tissue. **B:** Echo signals received from one beam direction into tissue.

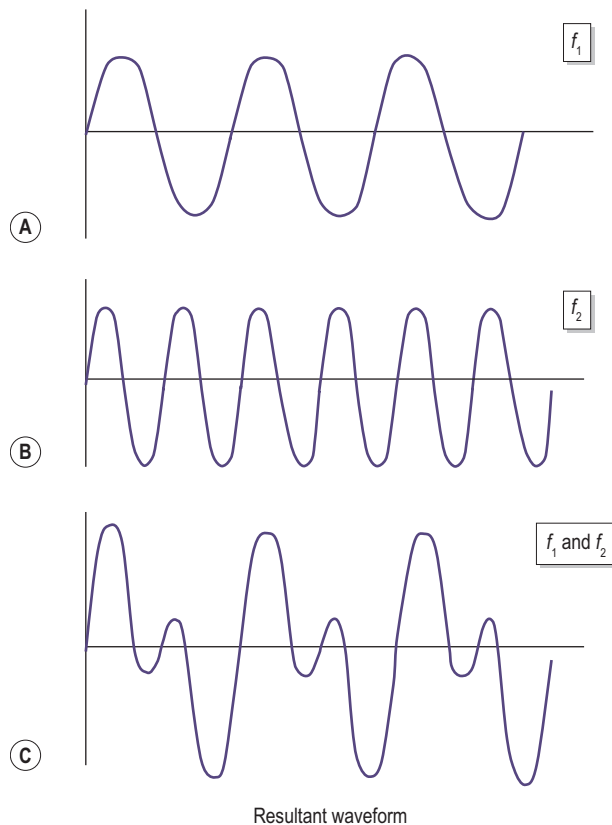
and some slightly bigger echoes from structures such as blood vessels or muscle fibres. The true speckle does not depict the very small tissue structures but the overall result of their echoes interfering. These tissue structures are smaller than can be imaged by ultrasound in the diagnostic frequency range. The speckle appearance of an image of tissue may help to identify the state of the tissue when the operator has a lot of experience with a particular machine. Comparison of speckle patterns between machines will probably not be valid as transducer design and the electronics influence the speckle appearance. In some applications tissue motion is measured by tracking the speckle pattern, for example in myocardial velocity imaging or elastography. The true speckle does not necessarily follow the tissue exactly but the echoes from the slightly bigger structures do and help to make the technique more robust.

With a sensitive B-mode scanner the speckle image of blood can be observed. The changing pattern relates to the motion of cells or groups of cells. The technique goes by the General Electric trade name of B-Flow.

## FOURIER COMPONENTS

In the discussion on interference, it was seen that when two waves or signals overlap the resultant wave pattern has a shape different from either of the two original waves. The simple case of two similar waves was considered giving constructive or destructive interference. However, if two or more waves or signals are considered of different frequencies and in and out of step by varying degrees, complex resultant waveforms can be produced (Fig. 1.9). The resultant waveform is said to be synthesised from the frequency components. The degree to which waves are in or out of step is the phase difference between the waves. Phase is important in the design of technology but it is not a concept that is of direct interest to the clinical user.

The opposite process of breaking a complex waveform down into its frequency components is of more direct interest in medical ultrasound. Then the complex waveform is said to be analysed into its frequency components. This is called frequency or Fourier analysis after its inventor. The more complex the waveform, the more Fourier components it has. A continuous wave of pure sinusoidal shape has one frequency component (Fourier component). A pulse has a range of frequency components. The shorter and sharper the pulse, the larger the range of frequency components; the range is known as the bandwidth of the pulse (Fig. 1.10). Each transducer



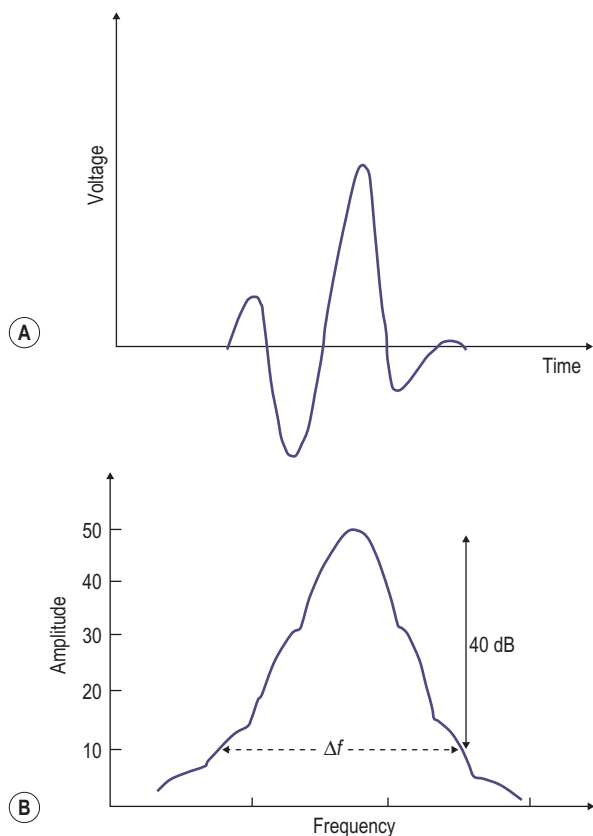
**Figure 1.9** Synthesis of a complex waveform by the addition of waves of different frequency. Frequency analysis of a complex waveform is the reverse process which reveals frequency components.

can only handle a specific range of frequencies, known as the bandwidth of the transducer. Ultrasound frequencies outside this range are attenuated and hence are lost to further processing. In Doppler blood flow techniques the complex Doppler signal is often analysed into its frequencies since they relate directly to the velocities of the blood cells at the site interrogated.

Some processes may remove frequencies from a waveform, e.g. absorption may remove the higher-frequency components of the signal. Other processes may add frequencies above or below the initial bandwidth of the transmitted ultrasound pulse, e.g. non-linear propagation or scattering at microbubbles. The frequencies that are added are often referred to as harmonics (above the upper bandwidth limit) or subharmonics (below the lower bandwidth limit) of the original signal.

## STANDING WAVES AND RESONANCE

When two waves of the same frequency travelling in opposite directions interfere, standing waves are formed. Examination of the wave pressure variations show alternate regions of high and low pressure amplitude oscillations that are not moving through the medium, i.e. a pattern of stationary nodes and anti-nodes has been formed (Fig. 1.11A). When a sound wave is reflected back and forth between two flat parallel surfaces in such a way that the sound travelling in opposite directions overlaps, standing waves are formed. If the separation of the surfaces is a whole number of half wavelengths, a marked increase in the pressure amplitude is observed (Fig. 1.11B). This is due to the waves adding constructively to give a resonance. When the separation of the surfaces is



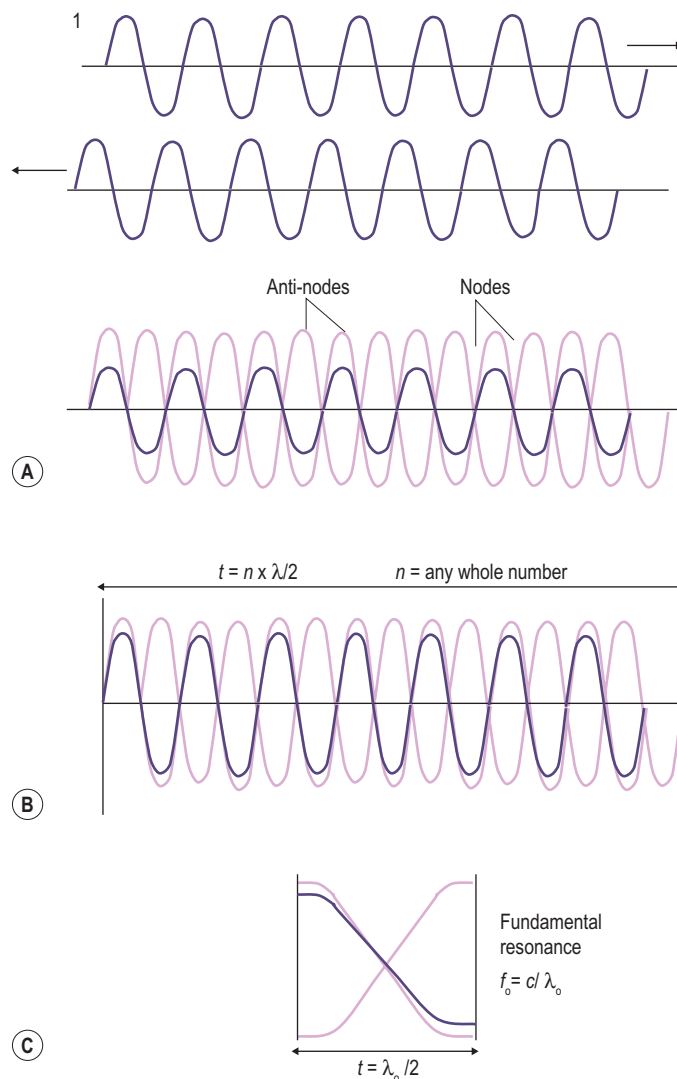
**Figure 1.10** The frequency spectrum of a pulsed signal. **A:** An echo pulse detected by a transducer. **B:** A plot of the amplitude (size) of the frequency components in the pulse after Fourier analysis.

one half-wavelength, a strong resonance occurs, known as the fundamental resonance (Fig. 1.11C).

At other separations equal to multiples of the half wavelength, other weaker resonances are known as harmonics. Conversely harmonics can also be seen when the separation is fixed and the wavelength is varied. These harmonics occur when the wavelength equals some multiple of the separation.

Piezoelectric transducer elements are made equal in thickness to one-half of the wavelength corresponding to their desired operating frequency to give efficient generation and detection of ultrasound. Continuous wave Doppler transducers have little or no damping and resonate at their operating frequency. Imaging and pulsed wave Doppler transducers have some damping which absorbs energy and spreads their sensitivity over a frequency range; i.e. they have a wider frequency response (bandwidth), which can be desirable though the damping reduces their sensitivity. When the size of an object has a special numerical relationship to the wavelength, a resonance will occur. An interesting example of resonance is exhibited when microbubbles of a particular size resonate in an ultrasound field. This resonance is exploited to improve the detection of these bubbles when they are injected into the bloodstream to act as imaging contrast agents (Chapter 6). By coincidence microbubbles, the size of red blood cells, resonate in the same MHz ultrasound frequency range as used in tissue imaging.

Standing waves and interference are of interest to the designer of transducers. They are two of the phenomena that contribute to the overall performance of a transducer and provide some insight into its complex operation. Given this complexity and scope for variation, it is always worth thoroughly assessing the performance of each transducer in clinical use.



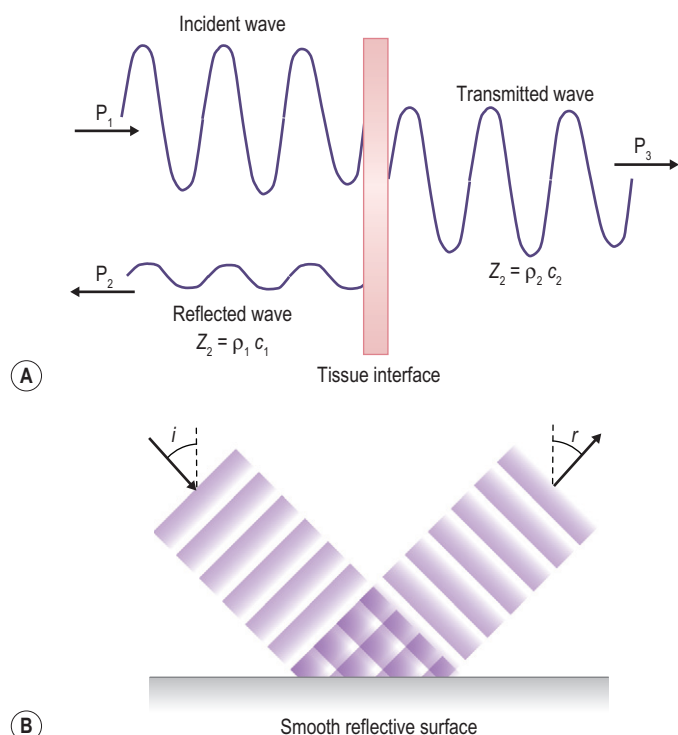
**Figure 1.11** **A:** The generation of a standing-wave pattern by two overlapping waves travelling in opposite directions. **B:** Standing waves produced by waves reflected within parallel surfaces. **C:** Fundamental resonance for surfaces separated by one half-wavelength.

#### Interference, frequencies and resonance

- Waves diffract and interfere.
- At high frequencies, and hence small wavelengths, diffraction can be controlled to produce directional beams.
- High frequencies produce narrow beams.
- Image speckle is produced by interference of echoes from small-scale structures within tissue. The speckle pattern is related to the structure but does not represent the actual structure.
- Complex waveforms (ultrasonic or electronic) can be broken down into frequency components. Frequency analysis is a very powerful technique for characterising signals.
- The frequency spectrum of a signal is important when 'harmonic imaging' and also Doppler blood flow detection are employed.
- Resonance can occur when ultrasound is reflected internally in a structure in which the dimensions of the structure are some multiple of the wavelength.
- Resonance is commonly encountered in transducer crystals and microbubbles.

## REFLECTION

Ultrasound is reflected when it strikes the boundary between two media where there is a change in density or compressibility or both. To be more exact, reflection occurs where there is a difference of acoustic impedance ( $Z$ ) between the media (Fig. 1.12A). The impedance is a measure of how readily tissue particles move under the influence of the wave pressure. The acoustic impedance of a medium equals the ratio of the pressure acting on the particles of the medium divided by the resulting velocity of motion of the particles. Therefore for tissues of different impedance a passing wave of particular pressure produces different velocities in each tissue. Note, the velocity of particle motion about the rest position is not the same as the velocity (speed) of the ultrasound waveform through the medium. For a wave in which the peaks and troughs of pressure lie in flat planes, plane waves, the acoustic impedance of a medium is equal to the density ( $\rho$ ) times the speed of sound in the medium, i.e.  $Z = \rho c$ . It is not surprising that reflection depends on quantities such as density and speed of sound, since the latter depends on the rigidity of the medium. In practice when imaging, echo size is often related roughly to the change in acoustic impedance at tissue boundaries. An approximate knowledge of tissue impedances is helpful in this respect (Table 1.1). Note that it is the change that is important – it does not matter whether there is an increase or decrease in impedance. The large changes in acoustic impedance at bone/soft tissue and gas/soft tissue boundaries are problematic since the transmitted pulse is then greatly reduced or even totally blocked in the case of gas by reflection at the boundary. The size of the echo in an image (i.e. the shade of grey) relates to the change in acoustic impedance at the interface producing it. Shades of grey are therefore related to the properties of tissues though signal processing in the scanner also plays an important part.



**Figure 1.12 Reflection.** **A:** Reflection at change in acoustic impedance between two media. **B:** Reflection at a smooth interface.

The unit of acoustic impedance is the Rayl, where 1 Rayl = 1 kg/m<sup>2</sup>s (units are often named after individuals, e.g. hertz, watt, pascal, rayleigh). As noted above, impedance is referred to in a semi-quantitative way; numerical values with units are rarely quoted except in some scientific studies, for example where attempts are made to characterise the ultrasonic properties of tissues. As can be seen from Table 1.1, the higher the density or stiffness of a material, the higher is its acoustic impedance.

It is instructive to consider the simple case of reflection of an incident ultrasound wave at a flat boundary between two media of impedances  $\rho_1 c_1$  and  $\rho_2 c_2$ . The magnitude of the echo amplitude is calculated using:

$$\text{Reflected amplitude} = \text{Incident amplitude} \times (\rho_1 c_1 - \rho_2 c_2) / (\rho_1 c_1 + \rho_2 c_2)$$

Table 1.2 gives a rough appreciation of echo sizes produced at different boundaries. Since properties of tissue are usually not accurately known and are highly dependent on the state of the tissue, significant differences from these values may exist in particular cases.

Reflection of ultrasound at a smooth surface is similar to light reflecting at a mirror and is sometimes referred to as specular reflection (Fig. 1.12B). Here it can be seen that the angle of incidence,  $i$ , is equal to the angle of reflection,  $r$ .

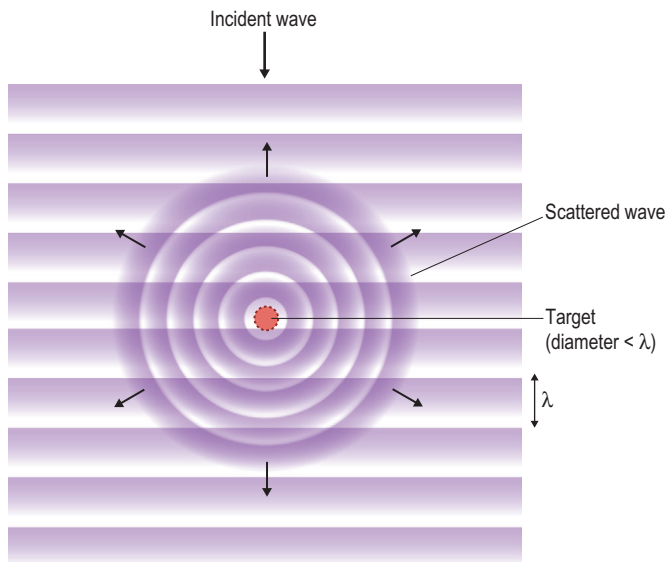
## SCATTERING

As an ultrasound wave travels through tissue it will probably interact with small tissue structures whose dimensions are similar to or less than a wavelength and whose impedances exhibit small variations. Some of the wave energy is then scattered in many directions (Fig. 1.13). Scattering is the process that provides most of the echo signals for both echo imaging and Doppler blood flow techniques. The closely packed scattering structures are very large in number and have a random distribution. Computer models of a wave interacting with such structures can predict a scattered wave with properties like those observed in practice. We noted earlier that the fluctuations in echo signals explain the fine speckled patterns detected from organ parenchyma and blood. Red cells in blood, singly or in the groups, are the scattering structures which produce the signals used in Doppler techniques. Reflection at a flat boundary, as discussed above, is a special case of scattering which occurs at smooth surfaces on which the irregularities are very much smaller than a wavelength. When the irregularities are comparable to the wavelength, they can no longer be ignored. This scattering at a non-smooth tissue interface enables it to be imaged more easily since echoes are detectable over a wide range of angles of incidence of the beam at the interface.

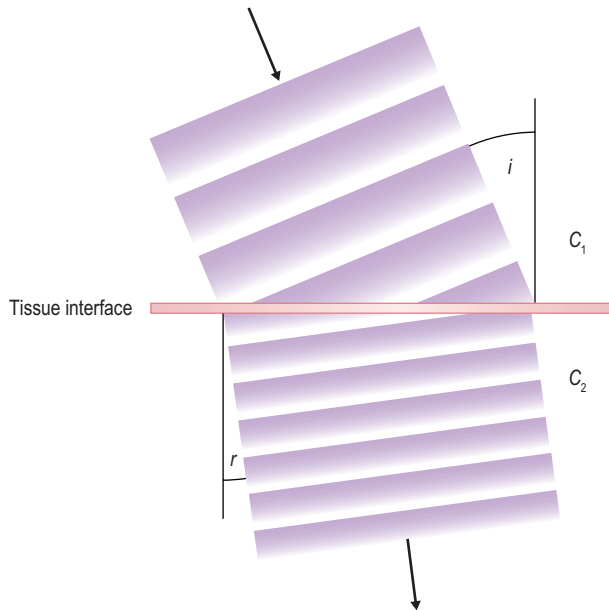
The ratio of the total ultrasound power,  $S$ , scattered in all directions by a target to the incident intensity,  $I$ , of the ultrasound beam is called the total scattering cross-section of the target. This ratio is used to compare the scattering powers of different tissues. Table 1.3

**Table 1.2 Percentage of incident energy reflected at tissue interfaces (perpendicular incidence at flat interface)**

Fat/muscle	1.08
Muscle/blood	0.07
Bone/fat	48.91
Fat/kidney	0.64
Lens/aqueous humour	1.04
Soft tissue/water	0.23
Soft tissue/air	99.90



**Figure 1.13 Wave scattering** at a target of dimensions much less than the wavelength.



**Figure 1.14 Deviation of an ultrasound beam** on striking at an angle the interface between two media of differing speed of sound.

**Table 1.3 Scattering within organs. Scattered signal level compared to typical soft tissue boundary (fat/muscle) echo level**

Reference boundary (fat/muscle)	Echo level	1.0
Liver parenchyma	Scatter level	0.032
Placenta	Scatter level	0.1
Kidney	Scatter level	0.01
Blood	Scatter level	0.001
Water	Scatter level	0.000

provides an idea of the scattering powers of some common tissues. The use of such scattering measurements has not been sufficiently discriminatory for tissue characterisation. However, the depiction of different levels of scattered echo signals in an image provides much of the information in the image.

## REFRACTION

The change in direction of a beam when it crosses a boundary between two media in which the speeds of sound are different is called 'refraction' (Fig. 1.14). If the angle of incidence is  $90^\circ$  there is no beam bending but at all other angles there is a change in direction. Refraction of light waves in optical components such as lenses and prisms is a common example. For an angle of incidence  $i$ , the new angle of the beam to the boundary on passing through it (the angle of refraction  $r$ ) is easily calculated using Snell's law:

$$\sin i / \sin r = c_1 / c_2 = \mu$$

where  $\mu$  is the refractive index.

Table 1.4 shows that deviations of a few degrees can occur at soft tissue boundaries. The results in ultrasonic imaging allow us to conclude that refraction is not a severe problem at soft tissue interfaces. An exception is at soft tissue/bone boundaries, where it can be a severe problem due to the big differences in speed of ultrasound between soft tissue and bone.

**Table 1.4 Beam deviation due to refraction at typical soft/tissue boundaries (angle of incidence  $30^\circ$ )**

Bone/Soft tissue	$19^\circ$
Muscle/Fat	$2.5^\circ$
Muscle/Blood	$0.5^\circ$
Muscle/Water	$1^\circ$
Lens/Aqueous humour	$2^\circ$

### Reflection, refraction and scattering

- An echo is generated at a change in acoustic impedance. The bigger the change, the bigger the echo. It does not matter if the change is an increase or decrease.
- The acoustic impedance of a tissue is related to the density and rigidity of the tissue.
- Bone and gas have acoustic impedances that are markedly different from those of soft tissue.
- Reflection is said to occur at surfaces of dimensions greater than the ultrasound wavelength and for smooth surfaces is analogous to light reflecting from a mirror or glass surface.
- Scattering occurs at small structures of acoustic impedance different from that of the surrounding medium. Scattering redirects the incident ultrasound over a wide range of angles, possibly through all angles in 3D.
- Very small tissue structures and blood cells are scattering centres of interest.
- Refraction occurs at a tissue interface where the velocity of ultrasound changes and the angle of incidence is not  $90^\circ$ .

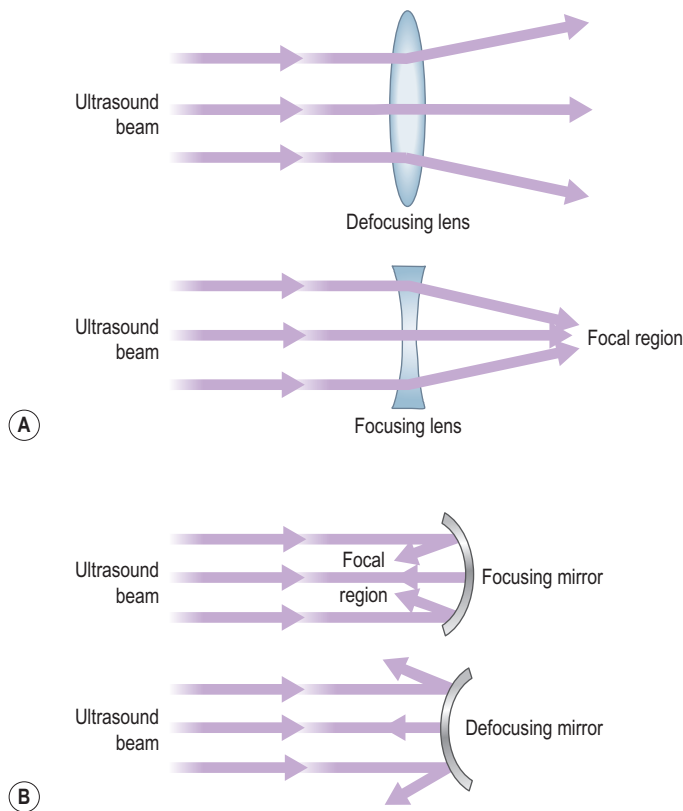
## LENSES AND MIRRORS

We often think of ultrasound behaving in a manner similar to that of light. This is not surprising since both are wave phenomena. As for light, lenses and mirrors can be constructed for ultrasound by exploiting reflection and refraction (Fig. 1.15). Lenses are designed by selecting material in which the velocity of ultrasound is different



**Table 1.5** Tissue thickness (in cm) to half intensity of ultrasound beam (-3 dB) for common clinical frequencies and materials

	1 MHz	3 MHz	5 MHz	10 MHz	20 MHz
Blood	17	8.5	3	2	1
Fat	5	2.5	1	0.5	0.25
Liver	3	1.5	0.5	-	-
Muscle	1.5	0.75	0.3	0.15	-
Bone	0.2	0.1	0.04	-	-
Polythene	0.6	0.3	0.12	0.6	0.03
Water	1360	340	54	14	3.4
Soft tissue (average)	4.3	2.1	0.86	0.43	0.21

**Figure 1.15** Lenses (A) and mirrors (B) for ultrasound.

from that of water or tissue. Mirror material has an acoustic impedance markedly different from water or tissue and hence results in very strong reflection. Lenses are most commonly found on the front face of transducers. Mirrors are rarely used.

## ABSORPTION AND ATTENUATION

As an ultrasound wave passes through tissue, its orderly vibrational energy is converted into random vibrational heat energy and hence the wave pressure amplitude reduces with distance travelled. This process is known as absorption. The higher the ultrasound frequency, the more rapidly is the amplitude reduced. Absorption rate also depends on the tissue involved.

In addition to absorption, other effects contribute to the total attenuation of the wave amplitude. These effects are reflection, scattering

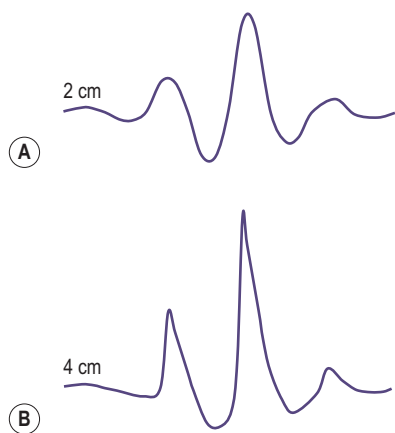
and beam divergence. The last two are frequency dependent, hence attenuation in tissue is highly dependent on frequency. Attenuation rises quickly over the diagnostic frequency range. Scanning equipment uses several techniques to compensate for attenuation and the operator should ensure that the appropriate controls have been optimised. This is further discussed in Chapter 3.

Table 1.5 presents attenuation for commonly encountered tissues. In the laboratory, attenuation is measured by noting the decrease in pressure amplitude as a wave passes through a known thickness of tissue. By taking the ratio of the input and output pressure amplitudes at either side of the slab of tissue, the drop in pressure amplitude can be quoted in dB/cm (attenuation coefficient). It is common practice to quote attenuation coefficients in dB/cm when imaging techniques are being considered, though in clinical application this would only be with special techniques related to tissue characterisation.

From Table 1.5 it can be seen that the attenuation in most soft tissues is similar and therefore an average value of 0.7 dB/cm/MHz can be used to compensate roughly for it (e.g. 2.1 dB/cm for 3 MHz ultrasound). The high attenuation of bone and calcified tissue is a particular problem. Later, in Chapter 3, controls will be discussed for attenuation compensation. It will be seen that compensation is applied by creating an image in which the average echo amplitude is made similar over the depth of tissue penetration. It will also be noted that some machines leave the compensation to a computer which assesses average signal level and applies compensation automatically (adaptively). This approach makes a lot of sense since it is in fact difficult to compensate for attenuation when the tissues being scanned are changing either due to their own motion or that of the scanning action of the transducer.

## NON-LINEAR PROPAGATION

Attenuation, discussed above, reduces the amplitude of ultrasound as it passes through tissue. It also reduces the high-frequency components in a pulse more than the lower ones and hence alters the shape of the pulse. This effect is rarely considered in clinical application. However, another phenomenon which alters the shape of a pulse and which is routinely exploited is called non-linear propagation. It is not significant for low-amplitude waves (e.g. 50 kPa) but important for high amplitude ones (e.g. 1 MPa). For waves of large pressure amplitude, the speed of sound is higher in regions where the pressure amplitude is positive than it is in the regions where the pressure is negative. The speed is different in the regions experiencing positive half-cycles than in negative half-cycles because the density of the medium changes with pressure. The effect is such that as the waveform passes through the tissue, the positive half-cycles catch up on the negative half-cycles resulting in distortion of



**Figure 1.16 A and B: Change of waveform shape** due to non-linear propagation in tissue. Note the high-frequency spikes indicating the production of harmonic frequencies.

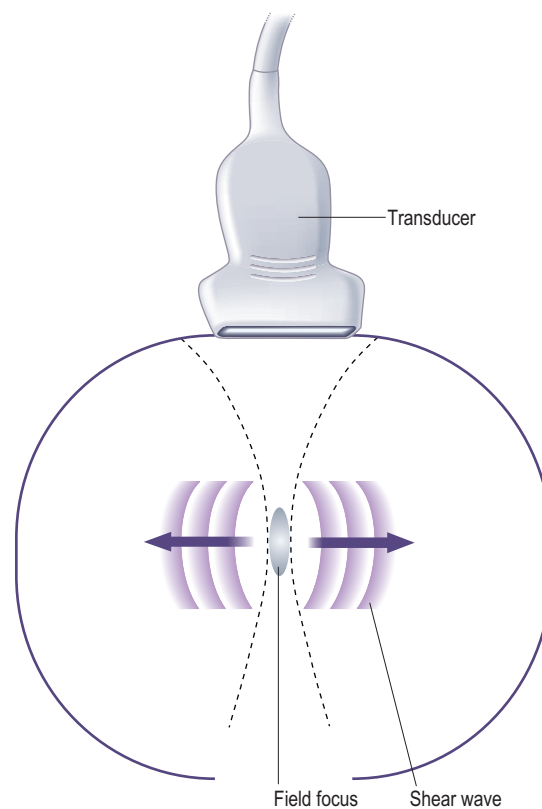
the waveform (Fig. 1.16). The greater the path travelled by the wave, the greater the distortion. The wave may become considerably distorted and have sharp discontinuities in its shape. The sinusoidal shape seen close to the transducer has then been replaced by a more saw-toothed shape, and the positive half-cycles can become particularly spiked.

The amount of non-linear distortion that occurs depends on the nature of the propagating medium as well as the wave parameters such as frequency and pressure amplitude. The amount of non-linear distortion is related to the molecular and structural properties of the medium in a complex way. However, the phenomenon has been very successfully exploited without the requirement to consider the differences between tissues. It is employed to produce narrower ultrasound beams rather than to provide information on the tissues through which the ultrasound passes.

Non-linear distortion is significant in the ultrasound methods used in medicine and has been exploited to improve images. The distortion is generated where the pressure is high, i.e. along the central axis of the ultrasound beam. Distortion is less in the weaker fringes or side-lobes of the beam, i.e. away from the central axis. By only using echoes produced by the non-linear transmitted pulse near the central axis, the scanning beam is effectively narrowed. It is possible to filter out these echoes since their shock-wave shape means that they contain a range of harmonic frequencies over and above those in the original transmitted pulse. It has been noted in the section on Fourier components that the shorter and sharper the shape, the bigger is the range of frequencies. These extra frequencies, which are generated as non-linear propagation distorts the pulse, are called harmonics. The technique utilising harmonics to produce narrower beams is known as harmonic imaging. The narrower beam improves the lateral resolution in the image. Harmonic imaging is very extensively used, indeed it is often the default mode which the machine is automatically programmed to activate when first switched on.

## TISSUE CHARACTERISATION AND ELASTOGRAPHY

When echo signals are collected there is often interest in saying more about the characteristics of the tissues that produced them. The most obvious characterisation would be to distinguish benign from malignant, but other distinctions may be attempted such as fatty from fibrotic liver or degree of calcification of arterial plaque. Parameters related to the phenomena described in previous sections may be measured to attempt tissue characterisation, e.g. speed



**Figure 1.17 The generation of shear waves** by the force at the focus of an ultrasound beam.

of sound, attenuation coefficient and scattering coefficient. Characterisation of tissue is often made difficult by the degradation of the ultrasound beam by fat and muscle between the transducer and the site of interest. Invasive imaging reduces this problem, for example by the use of catheter-based scanners to characterise blood vessel walls which are examined with a short beam path through blood. The problems are also reduced if there is a well-specified path between the transducer and the tissue of interest, e.g. through the eye to the retina or through a water-bath to skin lesions.

There is substantial development at the moment in the measurement of tissue elasticity by ultrasonic means. When a tissue is compressed by an external force, changes in its shape and size can be measured from the changes in position of the echo speckle pattern. From the size of the force and the tissue distortion, quantities can be calculated that relate to the elasticity of the tissue. This is analogous to palpation though the aim is to develop techniques that are more quantitative and can be applied to tissues deep within the body. It is known that the hardness of a mass can often be related to its pathology. Several different methods may be employed to apply a force to tissue, ranging from simply pushing the scanning transducer to using the beam radiation force to generate shear waves which spread out around the focus of a normal ultrasound beam. It was noted earlier that when the momentum of an ultrasound beam is interrupted by reflection or absorption at a target, the target experiences a radiation force. In the arrangement shown in Figure 1.17, a target tissue at the focus of the beam will experience a force impulse when subjected to an ultrasonic pulse. The resulting motion of the tissue produces a shear wave which travels sideways from the direction of the ultrasound beam, not unlike the ripple on a pond when a small pebble is dropped in. If several focal points are targeted at successive depths by successive transmitted pulses, the shear waves produced interfere with each other as they travel sideways and produce two plane shear waves travelling in opposite

directions at an angle to each other. This 'V'-shaped shear wave pattern resembles the sonic boom pressure wave generated by a jet aircraft travelling at the speed of sound. Labels such as 'radiation force' or 'supersonic' are used for these techniques.

Shear waves of MHz frequencies are attenuated very strongly in tissue and have not found application. However, shear waves of a few hundred Hz travel well through tissue and are the basis of the supersonic approach. The frequencies of the shear wave are determined by the pulsing rate of the ultrasound transducer. The speed of the shear waves and the wavelength are also very different from typical ultrasound waves as normally employed in ultrasound imaging (e.g. speed 1500 m/s and wavelength 0.5 mm for normal compression wave ultrasound and speed 1 m/s and wavelength 1 mm for shear waves in tissue). As the shear waves travel out from the focus (the shear wave source), the tissue moves under its influence, e.g. back and forth over typically 50 microns. The ultrasound scanner operating at a very high frame rate can image these small motions using signal processing methods similar to those used in colour Doppler imaging. From this image the speed of the shear wave at each point can be calculated. This speed depends on the elasticity (stiffness) of the tissue and hence can be used for diagnostic purposes. The results may be quoted in terms of the shear modulus in units of the pascal (Pa). Shear modulus is a scientific measure of the stiffness of the tissue. Considerable computation is required to provide the ultra-fast imaging and the elasticity images in real-time. Shear wave techniques are at quite an advanced state of development, and the first clinical machines are now available. Biological and clinical studies are in progress to establish their efficacy. The acoustic exposure of the patients is within FDA guidelines. In addition to elasticity, in the future measures of viscosity of tissue may be provided.

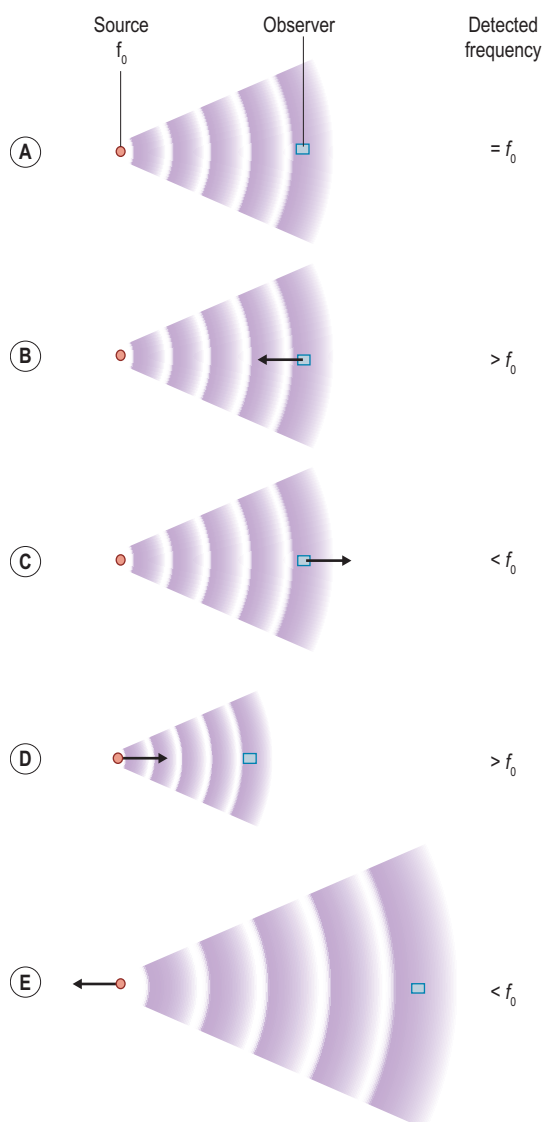
This field is normally called elastography or sonoelastography. It is a field that has been developed over the past 20 years and given the complex response of tissue to physical forces, this will continue both in terms of technological and clinical studies. That said, commercial equipment is now available for clinical evaluation.

#### Attenuation, propagation and harmonics

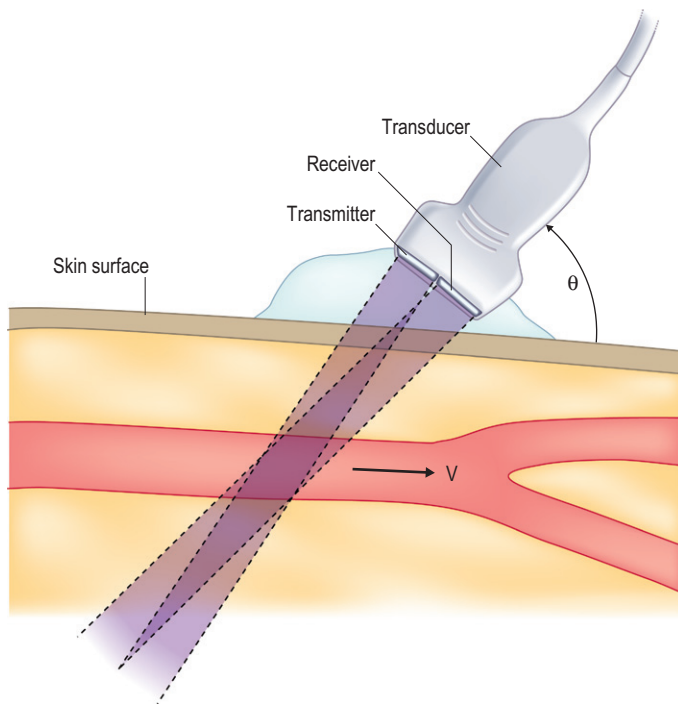
- Absorption occurs as ultrasound energy is converted into heat. It is highly frequency and tissue dependent.
- Attenuation is the reduction of beam intensity as it passes through tissue. It is the total effect of absorption, scattering, reflection, refraction and diffraction. It is highly frequency dependent. To obtain adequate echo signals from deep in tissue, the operator manipulates gain controls to compensate for attenuation.
- Non-linear propagation occurs as the wave pressure alters the properties of the tissue as it passes through it. It distorts the waveform which then contains additional frequency components – harmonics and subharmonics.
- Non-linear propagation occurs most near the axis of the beam where the wave amplitude is high relative to the sides of the beam. By only detecting the harmonics the beam is effectively made narrower, which gives better lateral resolution.
- Harmonics and subharmonics frequencies are also generated when ultrasound interacts with contrast microbubbles. Detection of these frequencies is exploited in the location of the bubbles in vivo.
- Many operators use harmonic imaging as the method of choice.
- Many attempts have been made to identify tissue types by measuring the acoustic properties of the tissue, e.g. attenuation, scattering and speed of sound. Some success has been achieved where there is well-defined anatomy but difficulties often arise where the ultrasound beam is distorted by overlying layers of fat and muscle.
- Present research into tissue characterisation seeks to develop elastography.

## DOPPLER EFFECT

When an ultrasound source and detector are at rest the frequency detected equals that transmitted provided the waveform is not distorted by the propagating tissue (Fig. 1.18A). The Doppler effect is the change in the observed frequency of a wave due to motion of the source of the wave or the observer. If the observer is moving towards a static source, an increase in frequency is observed since more wave cycles per second are encountered (Fig. 1.18B). Conversely for motion away from the source, fewer wave cycles are encountered per second and a decreased frequency is detected (Fig. 1.18C). The size of the Doppler shift is directly related to the size of the velocity. In another situation the source may move toward a static observer, the wavelengths are compressed as the source follows the wave and hence there is an increase in frequency (Fig. 1.18D). Motion of the source away from the observer stretches the wavelength and gives a decrease in frequency (Fig. 1.18E). Both of these cases of motion which give rise to changes in the observed frequency are in fact slightly different effects since in the first the wave is not altered and in the second it is compressed or stretched.



**Figure 1.18 A-E: The Doppler effect** in different situations relating to the relative motion of the source and observer.



**Figure 1.19** Examination of a blood vessel with a continuous wave Doppler transducer.

In medical ultrasound, a beam from an effectively static transducer is scattered back from tissues to produce echo signals (Fig. 1.19). The process of sending echoes back to the transducer is often called ‘backscatter’. Weak signals from blood are detected in this way and since the blood is moving the ultrasound has experienced a Doppler shift. In this situation if cells are moving towards the transducer, the total Doppler effect is produced by the cells moving through the waves plus the cells moving after the reflected ultrasound and hence compressing the echo signals. Motion of the cells towards the transducer produces an increase in ultrasound frequency due to this ‘double Doppler’ effect. Likewise, a reduction in frequency is produced by cells moving away from the transducer.

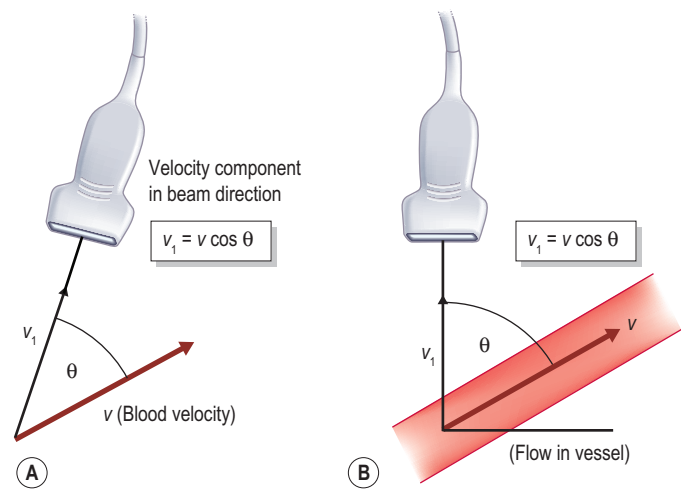
If the direction of the beam is at an angle to the direction of motion, it is the velocity component along the beam axis ( $v_1$ ) that is the relevant velocity when the Doppler effect is considered (Fig. 1.20A). Very often a simplified picture of blood flow is assumed in which the direction of flow is parallel to the walls of the blood vessel (Fig. 1.20B). This enables the beam/flow angle to be measured from an ultrasound image and hence the velocity to be calculated from the Doppler shift frequency.

A simple but useful Doppler equation links the Doppler shift, the transmitted frequency, the velocity and the angle between directions of sound propagation and motion:

$$f_d = 2vf_i \cos \theta / c \text{ or } v = f_d c / 2f_i \cos \theta$$

where  $f_d$  is the Doppler shift,  $f_i$  is the transmitted frequency,  $v$  is the velocity,  $\theta$  is the angle and  $c$  is the speed of ultrasound.

So far we have discussed blood cells moving with the same velocity giving rise to one Doppler shift frequency. Obviously in blood flow there can be many moving groups of cells with different velocities in a limited region in the beam. The limited region, which might typically be a few millimetres in diameter, is called the sample volume. Each group of cells gives rise to an echo signal with a particular Doppler shift and all of these signals combine to produce a complex Doppler signal from the blood. This complex signal can be analysed into a spectrum of frequencies by Fourier techniques. The frequency components in the spectrum can then be converted to velocities in the blood using the Doppler equation.



**Figure 1.20** A: Velocity component for angle between beam and motion directions. B: Measurement of velocity in a blood vessel.

#### Doppler techniques

- The Doppler effect is a very sensitive and accurate detector of motion. It is mostly used to study blood flow but it can also be employed to measure tissue motion.
- Users of Doppler techniques should be aware of the large effect of beam/motion angle on the velocity measurement.
- Most Doppler devices measure the velocity component along the ultrasound beam axis.
- Processes such as absorption, attenuation, scattering, refraction and non-linear propagation discussed for pulse-echo imaging also apply to Doppler methods.
- Doppler images and spectrograms are produced in real-time, making them very useful for the study of physiological function.

It is instructive to consider some numerical evaluations of this formula for values encountered in medical ultrasound. For example, if  $v = 20 \text{ cm/s}$ ,  $c = 154\,000 \text{ cm/s}$ ,  $\theta = 0^\circ$  ( $\cos \theta = 1$ ),  $f_i = 2 \times 10^6 \text{ Hz}$ , then the Doppler shift equals 519 Hz.

If the machine measures the Doppler shift and the beam angle, the velocity can be calculated and presented automatically on the screen. The Doppler shift is seen to be small, around one part in 1000, but electronics can measure it. The Doppler shift frequencies are also in the audible range, which means they can easily be studied simply by listening with earphones as well as using the spectral Doppler feature of machines.

It will be seen in Chapter 3 that the Doppler effect is exploited in various ways to present information on blood flow and tissue motion. A spectrum of velocities from a sample volume at a site of interest or colour images of velocities can be produced. Doppler methods probably provide the greatest amount of velocity information among all blood flow detection techniques.

## RESOLUTION

The detail that can be observed in ultrasound images or traces depends on the smallest change that can be presented by the ultrasound systems. The detail in a particular presentation usually is said to depend on one or more resolution. These resolutions depend on the shape of the ultrasound beam, the ultrasound pulse shape and the way they are transmitted. They are now listed since they apply to techniques described in later chapters.

Axial resolution in a pulse-echo image is the smallest separation of two targets lying along the beam axis for which individual echoes can be seen. Axial resolution depends on the length of the transmitted ultrasound pulse. Shorter pulses can be generated at higher ultrasound frequencies so higher (better) resolution is achievable at high frequencies.

In-plane (lateral) resolution is the smallest separation of two targets, placed side-by-side, for which separate echo spots are presented in an image. As the ultrasound beam sweeps across the scan plane, the echoes from the targets are detected one after the other if the targets are separated by more than the beam width. If the separation is less than the beam width, the echoes are detected simultaneously and they cannot be presented separately. The use of high frequencies and focusing enable generation of narrower beams and hence higher resolution is achieved. It is unfortunate that attenuation increases with frequency, limiting the upper values that can be used.

Out-of-plane (elevational, azimuthal) resolution is the minimum separation of two targets in the out-of-plane direction for which separate echoes can be detected. It is determined by the width of the ultrasound beam in the out-of-plane direction. This width is not always the same as the in-plane width. It is worth remembering when viewing an ultrasound image that the echoes have been collected from a tissue slice of finite width. Out-of-plane resolution is sometimes called slice-thickness resolution.

Contrast resolution is the smallest change in echo signal level that can be detected between regions in an image. This is most likely to be considered in terms of changes in shades of grey. Contrast resolution depends on the noise in the echo signal. The random fluctuations of noise may be from electronic sources in the scanner or from speckle resulting from the interference of echoes from small scattering structures in tissue. Averaging over a number of images can reduce noise, i.e. improve contrast resolution, but it slows up the imaging process. Contrast resolution is a somewhat neglected quantity but it is of major significance with regard to image quality.

Temporal resolution is the smallest separation in time for which two events can be identified separately. It depends on how quickly the information is refreshed. In imaging, temporal resolution depends on the rate at which images are produced, i.e. the frame rate. With physiological traces as encountered in Doppler techniques, the temporal resolution is related to the fast changes in the signal. We have seen earlier that the faster the changes occur in a trace, the greater is the range of frequency components in the trace. In other words, the greater the frequency bandwidth, the higher is the temporal resolution.

In Doppler imaging, different velocities are presented as different colours. The velocity resolution is the smallest change in velocity that can be detected. Likewise in a Doppler spectrogram, the velocity resolution is the smallest change in velocity that can be measured.

## APPENDIX

### Continuous wave ultrasound

- **Intensity spatial peak ( $I_{sp}$ ):** For a continuous wave beam, the intensity may be measured at the location of the maximum intensity in the beam (the spatial peak) to give the spatial peak intensity ( $I_{sp}$ ).
- **Intensity spatial average ( $I_{sa}$ ):** For a continuous wave beam, the intensity may be averaged across the beam to give the spatial average intensity ( $I_{sa}$ ).

### Pulsed wave ultrasound

For pulsed wave ultrasound, the intensity may be also averaged over the pulse length or the duration of the exposure time, i.e. temporal averaging may be carried out as well as spatial averaging.

- **Intensity spatial peak temporal average ( $I_{spta}$ ):** For a pulsed wave beam, the spatial peak intensity may be averaged over the duration of a sequence of several pulses to give the spatial peak, temporal average intensity ( $I_{spta}$ ).
- **Intensity spatial peak pulse average ( $I_{sppa}$ ):** For a pulsed wave beam, the spatial peak may be averaged over the pulse to give the spatial peak, pulse average intensity ( $I_{sppa}$ ).

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