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Edited by Peter Hoskins, Kevin Martin and Abigail Thrush
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Chapter 1

Introduction to B-mode imaging

Kevin Martin

The application of ultrasound to medical diagnosis has seen continuous development and growth over several decades. Early, primitive display modes, such as A-mode and static B-mode, borrowed from metallurgical testing and radar technologies of the time, have given way to high-performance, real-time imaging. Moving ultrasound images of babies in the womb are now familiar to most members of the public through personal experience of antenatal scanning or via television. Modern ultrasound systems do much more than produce images of unborn babies, however. Modern ultrasound systems are able to make detailed measurements of blood movements in blood vessels and tissues, visualize moving structures in 3D, and make measurements related to the stiffness of tissues.

Improvements in technology have been followed by widespread acceptance and use of ultrasound in medical diagnosis. Applications have progressed from simple measurements of anatomical dimensions, such as biparietal diameter, to detailed screening for fetal abnormalities, detection of subtle changes in tissue texture and detailed study of blood flow in arteries. In many areas, ultrasound is now chosen as the first line of investigation, before alternative imaging techniques.

This book describes the physics and technology of diagnostic ultrasound systems in use at the time of writing. The book may be divided into four sections; basic physics and B-mode imaging in Chapters 1–6; Doppler ultrasound in Chapters 7–10; quality assurance and safety in Chapters 11–12, and recent technology in Chapters 13–15. This chapter covers the very basic concepts involved in B-mode imaging.

Basic principles of ultrasound image formation

We begin the explanation of ultrasound image formation with a description of a B-mode image and the basic principles of its formation. In essence, these principles are still used in modern B-mode systems, although they may be used within more complex arrangements designed to enhance performance.

A B-mode image is a cross-sectional image representing tissues and organ boundaries within the body (Figure 1.1). It is constructed from echoes, which are generated by reflection of ultrasound waves at tissue boundaries, and scattering from small irregularities within tissues. Each echo is displayed at a point in the image, which corresponds to the relative position of its origin within the body cross section, resulting in a scaled map of echo-producing features. The brightness of the image at each point is related to the strength or amplitude of the echo, giving rise to the term B-mode (brightness mode).

Usually, the B-mode image bears a close resemblance to the anatomy, which might be seen by eye, if the body could be cut through in the same plane. Abnormal

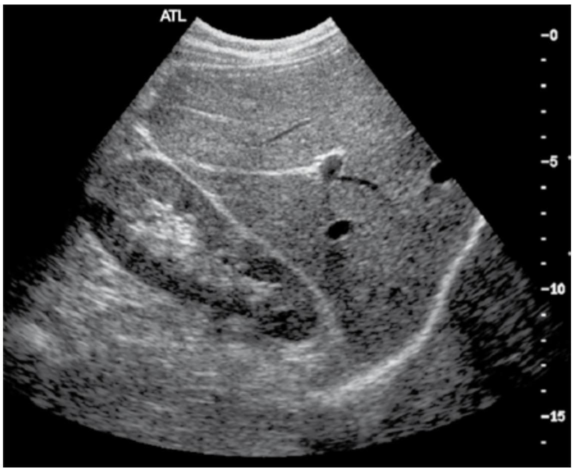

Fig. 1.1 An example of a B-mode image showing reflections from organ and blood vessel boundaries and scattering from tissues.

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anatomical boundaries and alterations in the scattering behaviour of tissues can be used to indicate pathology.

To form a B-mode image, a source of ultrasound, the transducer, is placed in contact with the skin and short bursts or pulses of ultrasound are sent into the patient. These are directed along narrow beam-shaped paths. As the pulses travel into the tissues of the body, they are reflected and scattered, generating echoes, some of which travel back to the transducer, where they are detected. These echoes are used to form the image.

To display each echo in a position corresponding to that of the interface or feature (known as a target) that caused it, the B-mode system needs two pieces of information. These are

(1) the range (distance) of the target from the transducer and
(2) the direction of the target from the active part of the transducer, i.e. the position and orientation of the ultrasound beam.

Echo ranging

The range of the target from the transducer is measured using the pulse–echo principle. The same principle is used in echo-sounding equipment in boats to measure the depth of water. Figure 1.2 illustrates the measurement of water depth using the pulse–echo principle. Here, the transducer transmits a short burst or pulse of ultrasound, which travels through water to the seabed below, where it is reflected, i.e. produces an echo. The echo travels back through the water to the transducer, where it is detected. The distance to the seabed can be worked out, if the speed of sound in water is known and the time between the pulse leaving the transducer and the echo being detected, the 'go and return time', is measured.

To measure the go and return time, the transducer transmits a pulse of ultrasound at the same time as a clock is started ($t = 0$). If the speed of sound in water is c and the depth is d, then the pulse reaches the seabed at time $t = d/c$. The returning echo also travels at speed c and takes a further time d/c to reach the transducer, where it is detected. Hence, the echo arrives back at the transducer after a total go and return time $t = 2d/c$. Rearranging this equation, the depth d can be calculated from $d = ct/2$. Thus, the system calculates the target range d by measuring the arrival time t of an echo, assuming a fixed value for the speed of sound c (usually 1540 m s^{-1} for human tissues).

In the above example, only one reflecting surface was considered, i.e. the interface between the water and

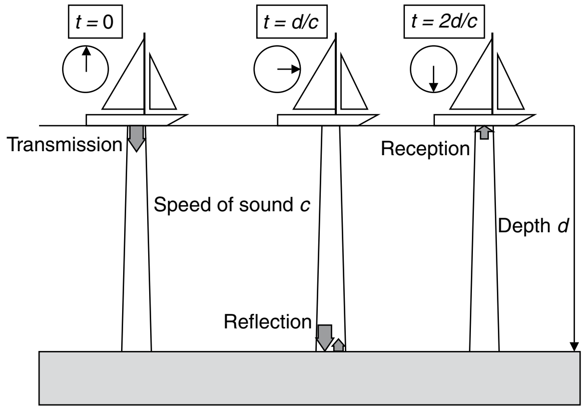

Fig. 1.2 Measurement of water depth using the pulse–echo principle. The depth is worked out by measuring the time from transmission of the pulse to reception of the echo. The speed of sound must be known.

the seabed. The water contained no other interfaces or irregularities, which might generate additional echoes. When a pulse travels through the tissues of the body, it encounters many interfaces and scatterers, all of which generate echoes. After transmission of the short pulse, the transducer operates in receive mode, effectively listening for echoes. These begin to return immediately from targets close to the transducer, followed by echoes from greater and greater depths, in a continuous series, to the maximum depth of interest. This is known as the pulse–echo sequence.

Image formation

The 2D B-mode image is formed from a large number of B-mode lines, where each line in the image is produced by a pulse–echo sequence. In early B-mode systems, the brightness display of these echoes was generated as follows.

As the transducer transmits the pulse, a display spot begins to travel down the screen from a point corresponding to the position of the transducer, in a direction corresponding to the path of the pulse (the ultrasound beam). Echoes from targets near the transducer return first and increase the brightness of the spot. Further echoes, from increasing depths, return at increasing times after transmission as the spot travels down the screen. Hence, the distance down the display at which each echo is displayed is related to its depth below the transducer. The rate at which the display spot travels down the screen determines the scale of the image. A rapidly moving spot produces a magnified image.

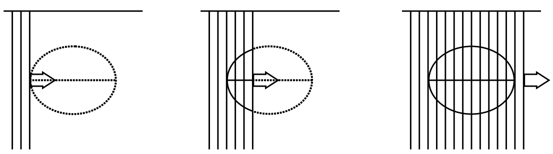

Fig. 1.3 Formation of a 2D B-mode image. The image is built up line by line as the beam is stepped along the transducer array.

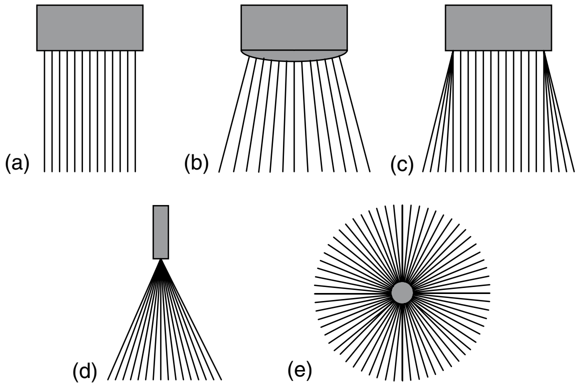

Fig. 1.4 Scan line arrangements for the most common B-mode formats. These are (a) linear, (b) curvilinear, (c) trapezoidal, (d) sector and (e) radial.

The pulse–echo sequence, described above, resulted in the display of one line of information on the B-mode image. A complete B-mode image, such as that in Figure 1.1, is made up typically of 100 or more B-mode lines.

Let us consider a linear array probe, as described in Chapter 3, where the image is formed as illustrated in Figure 1.3. During the first pulse–echo sequence, an image line is formed, say on the left of the display. The active area of the transducer, and hence the beam, is then moved along the array to the adjacent beam position. Here a new pulse–echo sequence produces a new image line of echoes, with a position on the display corresponding to that of the new beam. The beam is progressively stepped along the array with a new pulse–echo sequence generating a new image line at each position.

One complete sweep may take perhaps 1/30th of a second. This would mean that 30 complete images could be formed in 1 s, allowing real-time display of the B-mode image. That is, the image is displayed with negligible delay as the information is acquired, rather than recorded and then viewed, as with a radiograph or CT scan.

B-mode formats

The B-mode image, just described, was produced by a linear transducer array, i.e. a large number of small transducer elements arranged in a straight line (see Chapter 3). The ultrasound beams, and hence the B-mode lines, were all perpendicular to the line of transducer elements, and hence parallel to each other (Figure 1.4a). The resulting rectangular field of view is useful in applications, where there is a need to image superficial areas of the body at the same time as organs at a deeper level.

Other scan formats are often used for other applications. For instance, a curvilinear transducer (Figure 1.4b) gives a wide field of view near the transducer and an even wider field at deeper levels. This is also achieved by the trapezoidal field of view (Figure 1.4c). Curvilinear and trapezoidal fields of view are widely used in obstetric scanning to allow imaging of more superficial targets, such as the placenta, while giving the greatest coverage at the depth of the baby. The sector field of view (Figure 1.4d) is preferred for imaging of the heart, where access is normally through a narrow acoustic window between the ribs. In the sector format, all the B-mode lines are close together near the transducer and pass through the narrow gap, but diverge after that to give a wide field of view at the depth of the heart.

Transducers designed to be used internally, such as intravascular or rectal probes, may use the radial format (Figure 1.4e) as well as sector and linear fields of view. The radial beam distribution is similar to that of beams of light from a lighthouse. This format may be obtained by rotating a single element transducer on the end of a catheter or rigid tube, which can be inserted into the body. Hence, the B-mode lines all radiate out from the centre of the field of view.

Chapter 2

Physics

Kevin Martin and Kumar Ramnarine

Introduction

Ultrasound is a high-frequency sound wave, which can be used to form images of internal body organs, as described briefly in the previous chapter. Ultrasound travels through the tissues of the body in a way which makes it possible to form useful images using relatively simple techniques, as described in Chapter 1. However, the image-formation process includes some approximations, which give rise to imperfections and limitations in the imaging system. In order to be able to use diagnostic ultrasound systems effectively and to be able to distinguish imperfections in the image from genuine diagnostic information, the user must have an appreciation of the basic principles of ultrasound propagation in tissue.

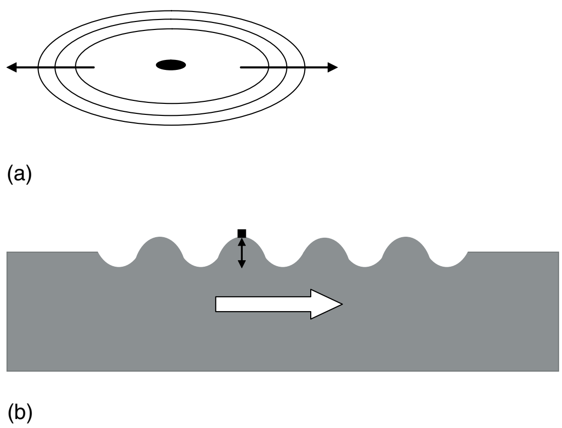

Fig. 2.1 Waves on the surface of a pond: (a) waves on the surface of a pond travel out from the point of entry of a stone; (b) only the disturbance travels across the pond. The water surface simply goes up and down.

Waves

Transverse waves

A wave is a disturbance with a regularly repeating pattern, which travels from one point to another. A simple and familiar example is a wave on the surface of a pond caused by a stone being thrown into the water (Figure 2.1a). Here, water displaced by the stone causes a local change in the height of the water, which causes a change in height in the water immediately adjacent to it and so on. Hence a wave travels out from the point of entry of the stone. An important aspect of the nature of this wave is that it is only the disturbance which travels across the pond, and not the water. The surface of the water at each point in the pond, as shown by a floating object (Figure 2.1b), simply goes up and down like a weight on the end of a spring, giving rise to the oscillating nature of the wave. Energy is transported across the pond from the stone to the shore. This type of wave on the surface of water is described as a transverse wave because the local movement of the water surface is at 90° (transverse) to the direction of travel.

Sound waves

The sound waves used to form medical images are longitudinal waves, which propagate (travel) through a physical medium (usually tissue or liquid). Here, the particles of the medium oscillate backwards and forwards along the direction of propagation of the wave (see Figure 2.2). Where particles in adjacent regions have moved towards each other, a region of compression (increased pressure) results, but where particles have moved apart, a region of rarefaction (reduced pressure) results. As in the transverse wave case, there is no net movement of the medium. Only the disturbance and its associated energy are transported.

The most familiar sound waves are those that travel in air from a source of sound, e.g. a musical instrument

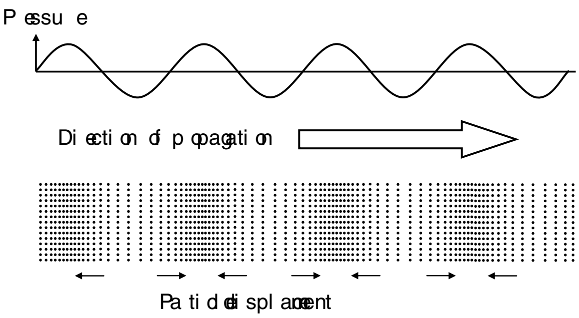

Fig. 2.2 In a longitudinal wave, particle motion is aligned with the direction of travel, resulting in bands of high and low pressure.

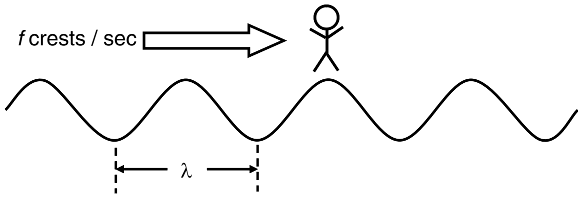

Fig. 2.3 The frequency f of a wave is the number of wave crests passing a given point per second. The wavelength λ is the distance between wave crests.

or a bell, to the human ear. The surface of a bell vibrates when it is struck. The oscillating motion of the surface pushes and pulls against the air molecules adjacent to it. Neighbouring air molecules are then set in motion, which displace their neighbours and so the disturbance travels through the air as a sound wave. When the sound wave reaches the listener's ear, it causes the eardrum to vibrate, giving the sensation of sound. Energy from the bell is transported by the wave to the eardrum, causing it to move.

Frequency, speed and wavelength

Frequency

When the bell above is struck, its surface vibrates backwards and forwards at a certain frequency (number of times per second). An observer listening to the sound at any point nearby will detect the same number of vibrations per second. The frequency of the wave is the number of oscillations or wave crests passing a stationary observer per second (Figure 2.3) and is determined by the source of the sound wave. Frequency is normally given the symbol f and has units of hertz (1 Hz = 1 cycle per second). Sound waves with frequencies in the approximate range 20 Hz to 20 kHz can be detected by the human ear. Sound waves with frequencies above approximately 20 kHz cannot be heard and are referred to as ultrasound waves.

Speed

As will be shown in more detail later, the speed at which a sound wave travels is determined by the medium in which it is travelling. The speed of sound is normally given the symbol c and has units of m s^{-1} (metres per second). Examples are the speed of sound in air (330 m s^{-1}) and water (1480 m s^{-1}).

Wavelength

The wavelength of a wave is the distance between consecutive wave crests or other similar points on the wave, as illustrated in Figure 2.3. Wavelength is normally given the symbol λ (lambda) and has units of metres or millimetres.

A wave whose crests are λ metres apart and pass an observer at a rate of f per second must be travelling at a speed of $f \times \lambda$ metres per second. That is, the speed of sound $c = f\lambda$. However, it is more accurate physically to say that a wave from a source of frequency f, travelling through a medium in which the speed of sound is c, has a wavelength λ, where $\lambda = c/f$.

For example, a sound wave from a 30 kHz source travelling through water ($c \approx 1500$ m s^{-1}) has a wavelength of 50 mm, whereas a wave from the same source travelling through air ($c = 330$ m s^{-1}) has a wavelength of about 10 mm.

Phase

As a sound wave passes through a medium, the particles are displaced backwards and forwards from their rest positions in a repeating cycle. The pattern of displacement of the particles with time can often be described by a sine wave (Figure 2.4a). This pattern of displacement is as would be seen in the height of a rotating bicycle pedal when viewed from behind the bicycle. A complete cycle of the pedal height corresponds to a complete 360° rotation. The height of the pedal at any point in the cycle is related to the angle of the pedal when viewed from the side. The phase of the pedal is its position within such a cycle of rotation and is measured in degrees. For example if a position of horizontal (zero height) and to the rear is defined as a phase of 0°, a phase of 90° will correspond to the pedal being in the vertical position where its height is at a maximum. At 180°, the pedal is horizontal and forwards with a

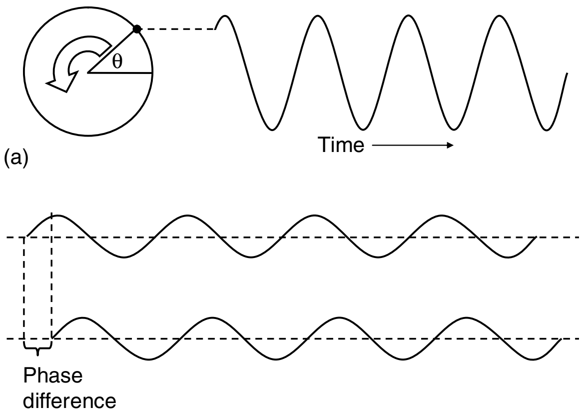

Fig. 2.4 (a) Phase describes the position within a cycle of oscillation and is measured in degrees. (b) Two waves of the same frequency and amplitude can be compared in terms of their phase difference.

height of zero. The height reaches its minimum value at 270° when the pedal is vertically down.

Two waves of the same frequency may differ in terms of their phase and can be compared in terms of their phase difference, measured in degrees (Figure 2.4b). Phase difference is an important concept when waves are added together, as described later in this chapter.

Pressure, intensity and power

As explained earlier, a sound wave passing through a medium causes the particles of the medium to oscillate back and forth along the direction of propagation (i.e. longitudinally). The maximum distance moved by a particle from its normal rest position is a measure of the amplitude (or strength) of the wave. This is referred to as the displacement amplitude. The longitudinal motion of the particles results in regions of compression and rarefaction so that at each point in the medium the pressure oscillates between maximum and minimum values as the wave passes. The difference between this actual pressure and the normal rest pressure in the medium is called the excess pressure, p, which is measured in pascals (Pa), where 1 Pa equals 1 N m^{-2} (the newton N is a measure of force). When the medium is compressed, the excess pressure is positive. When the medium undergoes rarefaction, the pressure is less than the normal rest pressure, and so the excess pressure is negative. The amplitude of the wave may also be described by the peak excess pressure, the maximum value during the passage of a wave. Excess pressure is commonly referred to simply as the pressure in the wave.

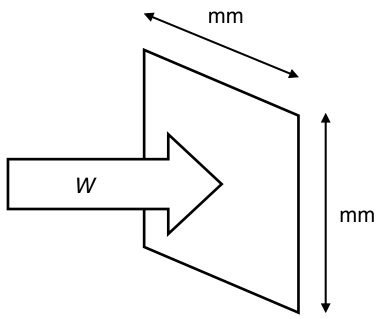

Fig. 2.5 Intensity is the power W flowing through unit area e.g. 1 W m^{-2}.

As the sound wave passes through the medium, it transports energy from the source into the medium. The rate at which this ultrasound energy is produced by the source is given by the ultrasound power. Energy is measured in joules (J) and power is measured in watts (W) where 1 W = 1 J s^{-1}.

The ultrasound produced by the source travels through the tissues of the body along an ultrasound beam, and the associated power is distributed across the beam. As will be seen later in this chapter, the power is not distributed evenly across the beam, but may be more concentrated or intense near the centre. The intensity is a measure of the amount of power flowing through an area of the beam cross section. Intensity is defined as the power flowing through unit area presented at 90° to the direction of propagation (Figure 2.5). Intensity I is measured in W m^{-2} or mW cm^{-2}.

As one might expect intuitively, the intensity associated with a wave increases with the pressure amplitude of the wave. In fact intensity I is proportional to p^2.

Speed of sound

The speed of propagation of a sound wave is determined by the medium it is travelling in. In gases (e.g. air) the speed of sound is relatively low in relation to values in liquids, which in turn tend to be lower than values in solids. The material properties which determine the speed of sound are density and stiffness. Density is a measure of the weight of a material for a given volume. For example, a 5 cm cube of steel weighs much more than a 5 cm cube of wood because steel has a higher density. Density is normally given

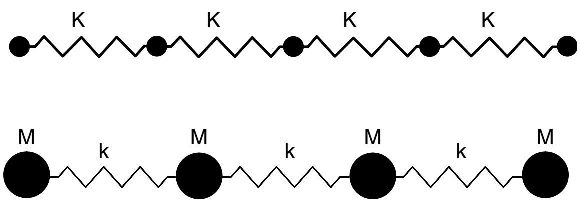

Fig. 2.6 The speed of sound in a medium is determined by its density and stiffness, which can be modelled by a series of masses and springs.

the symbol ρ (rho) and is measured in units of kg m^{-3}. Stiffness is a measure of how well a material resists being deformed when it is squeezed. This is given by the pressure required to change its thickness by a given fraction. Stiffness is usually denoted by the symbol k (units of Pa).

A simple picture of how the density and stiffness of a material determine its speed of sound can be obtained from the model shown in Figure 2.6, which consists of two lines of weights, or more correctly masses, connected by springs. The small masses (m) model a material of low density and the large masses (M) a material of high density. In the two models shown, the small masses are linked by springs of high stiffness K and the large masses by springs of low stiffness k. A longitudinal wave can be propagated along the row of small masses (m) by giving the first mass a momentary push to the right. This movement is coupled to the second small mass by a stiff spring causing it to accelerate quickly to the right and pass on the movement to the third mass, and so on. As the masses are light (low density), they can be accelerated quickly by the stiff springs (high stiffness) and the disturbance travels rapidly.

In the second case, a momentary movement of the first large mass M to the right is coupled to the second mass by a weak spring (low stiffness). The second large mass will accelerate relatively slowly in response to the small force from the weak spring. Its neighbours to the right also respond slowly so that the disturbance travels relatively slowly.

Hence, low density and high stiffness lead to high speed of sound whereas high density and low stiffness lead to low speed of sound. Mathematically this is expressed in the following equation:

$$\text{Speed of sound } c = \sqrt{k/\rho}$$

Table 2.1 Speed of sound in human tissues and liquids (from Duck 1990).

Material	c (m s^{-1})
Liver	1578
Kidney	1560
Amniotic fluid	1534
Fat	1430
Average tissue	1540
Water	1480
Bone	3190–3406
Air	333

Although gases have low density, they have very low stiffness (high compressibility), leading to relatively low speed of sound compared to liquids and solids.

Table 2.1 shows typical values for the speed of sound in various materials, including a number of different kinds of human tissue. The most important point to note from this table is that the values for the speed of sound in human soft tissues are rather similar. In fact they are sufficiently similar that the B-mode image-forming process can assume a single, average value of 1540 m s^{-1} without introducing significant errors or distortions in the image. All the values shown (with the exception of fat) are within 5% of this average value and are not much different from the value in water. The speed of sound in air is much lower because of its low stiffness, and that in bone is much higher because of its high stiffness.

Frequencies and wavelengths used in diagnosis

The ultrasound frequencies used most commonly in medical diagnosis are in the range 2–15 MHz, although frequencies up to 40 MHz may be used in special applications and in research. The wavelengths in tissue which result from these frequencies can be calculated using the equation given earlier, which relates wavelength λ to the frequency f and speed c of a wave:

$$\lambda = \frac{c}{f}$$

Assuming the average speed of sound in soft tissues of 1540 m s^{-1}, values of λ at diagnostic frequencies are as shown in Table 2.2.

Table 2.2 Wavelengths used in diagnosis.

f (MHz)	λ (mm)
2	0.77
5	0.31
10	0.15
15	0.1

The wavelengths in soft tissues which result from these frequencies are within the range 0.1–1 mm. As will be seen later in this chapter and in Chapter 5, the wavelength of the ultrasound wave has an important influence on the ability of the imaging system to resolve fine anatomical detail. Short wavelengths give rise to improved resolution, i.e. the ability to show closely spaced targets separately in the image.

Reflection of ultrasound waves

In Chapter 1, a B-mode image was described as being constructed from echoes, which are generated by reflections of ultrasound waves at tissue boundaries and by scattering from small irregularities within tissue. Reflections occur at tissue boundaries where there is a change in acoustic impedance (see below). When an ultrasound wave travelling through one type of tissue encounters an interface with a tissue with different acoustic impedance, some of its energy is reflected back towards the source of the wave, while the remainder is transmitted into the second tissue.

Acoustic impedance

The acoustic impedance of a medium z is a measure of the response of the particles of the medium in terms of their velocity, to a wave of a given pressure. Acoustic impedance $z = p/v$, where p is the local pressure and v is the local particle velocity. It is analogous to electrical impedance (or resistance R), which is the ratio of the voltage (V) applied to an electrical component (the electrical driving force or pressure) to the resulting electrical current (I) which passes through it (the response), as expressed in Ohm's law: $R = V/I$.

The acoustic impedance of a medium is again determined by its density (ρ) and stiffness (k). It can be explained in more detail, as with the speed of sound, by modelling the medium as a row of small or large masses (m) and (M) linked by weak or stiff springs (k) and (K) as shown in Figure 2.7.

In this case, however, the small masses m are linked by weak springs k, modelling a material with low density and low stiffness. The large masses M are linked by stiff springs K, modelling a material with high density and stiffness.

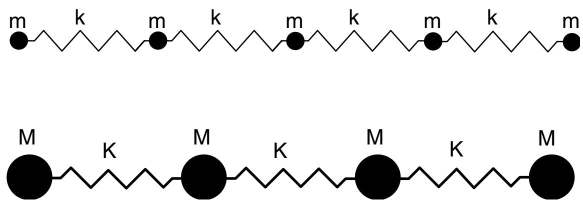

Fig. 2.7 The acoustic impedance of a medium is determined by its density and stiffness, which can be modelled by a series of masses and springs.

If a given pressure (due to a passing wave) is applied momentarily to the first small mass m, the mass is easily accelerated to the right (reaching increased velocity) and its movement encounters little opposing force from the weak spring k. This material has low acoustic impedance, as particle movements within it (in terms of velocity) in response to a given pressure are relatively large. In the second case, the larger masses M accelerate less in response to the applied pressure (reaching lower velocity) and their movements are further resisted by the stiff springs. Particle velocity (the response) in this material is lower for a given applied pressure and it has higher acoustic impedance. The acoustic impedance z of a material is given by:

$$z = \sqrt{\rho k}$$

By combining this equation with that for the speed of sound given earlier, it can be shown also that:

$$z = \rho c$$

Acoustic impedance z has units of kg m^{-2} s^{-1}, but the term rayl (after Lord Rayleigh) is often used to express this unit.

Table 2.3 gives values of z for some common types of human tissue, air and water. Table 2.3 shows that values for z in most human soft tissues are very similar. For air, which has low density and low stiffness, z is very small. For bone, which has high density and high stiffness, z is much higher.

Reflection

When a sound wave travelling through one medium meets an interface with a second medium of different acoustic impedance, some of the wave is transmitted

Table 2.3 Values of acoustic impedance.

Material	z (kg m^{-2} s^{-1})
Liver	1.66×10^6
Kidney	1.64×10^6
Blood	1.67×10^6
Fat	1.33×10^6
Water	1.48×10^6
Air	430
Bone	6.47×10^6

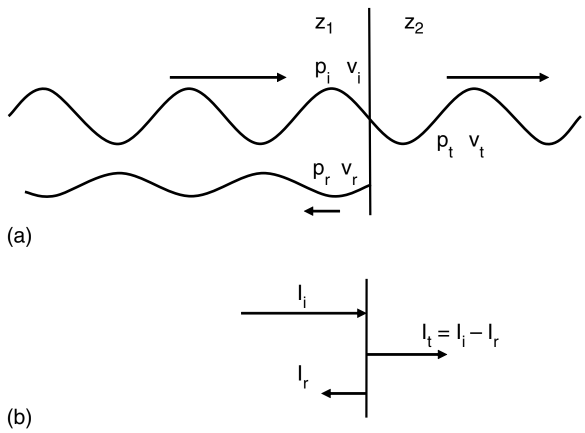

Fig. 2.8 (a) Total particle pressure and velocity cannot change abruptly across an interface. So a reflected wave is formed when there is a change in acoustic impedance. (b) The intensity transmitted across an interface is the incident intensity minus that reflected.

into the second medium and some is reflected back into the first medium. The amplitudes of the transmitted and reflected waves depend on the change in acoustic impedance. Figure 2.8a shows a sound wave travelling through a medium with acoustic impedance z_1, incident on an interface with a second medium with acoustic impedance z_2. The acoustic impedance changes abruptly at the interface. However, the motion of the particles of the medium, and hence their pressure and velocity, must be continuous across the interface to avoid disruption of the medium. To achieve this, the total wave pressure and velocity at the interface in medium 1 must equal those in medium 2 near to the interface. That is $p_i + p_r = p_t$ and $v_i + v_r = v_t$.

From this condition it can be shown that:

$$\frac{p_r}{p_i} = \frac{z_2 - z_1}{z_2 + z_1}$$

Table 2.4 Amplitude reflection coefficients of interfaces.

Interface	R_A
Liver–kidney	0.006
Kidney–spleen	0.003
Blood–kidney	0.009
Liver–fat	0.11
Liver–bone	0.59
Liver–air	0.9995

where p_i and p_r are the pressure amplitudes of the incident and reflected waves respectively near the interface.

This ratio of reflected to incident pressure is commonly referred to as the amplitude reflection coefficient R_A. It is very important to ultrasound image formation as it determines the amplitude of echoes produced at boundaries between different types of tissue.

Table 2.4 shows values of amplitude reflection coefficient for some interfaces that might be encountered in the body. For most soft tissue to soft tissue interfaces, the amplitude reflection coefficient is less than 0.01 (1%). This is another important characteristic for ultrasound imaging as it means that most of the pulse energy at soft tissue interfaces is transmitted on to produce further echoes at deeper interfaces. The amplitude reflection coefficient at a tissue–fat interface is about 10% due to the low speed of sound in fat. At an interface between soft tissue and air, as might be encountered within the lungs or gas pockets in the gut, the reflection coefficient is 0.999 (99.9%), so no further useful echoes can be obtained from beyond such an interface. For this reason, it is important to exclude air from between the ultrasound source (the transducer) and the patient's skin to ensure effective transmission of ultrasound. At an interface between soft tissue and bone, the amplitude reflection coefficient is approximately 0.5 (50%), making it difficult also to obtain echoes from beyond structures such as ribs. Note that the reflection coefficient is not related to the frequency of the wave; it is determined only by the change in z at the interface between the two media.

The intensity reflection coefficient describes the ratio of the intensities of the reflected (I_r) and incident waves (I_i). As intensity is proportional to pressure squared, the intensity reflection coefficient R_i is given by:

$$\frac{I_r}{I_i} = R_i = R_A^2 = \left(\frac{z_2 - z_1}{z_2 - z_1}\right)^2$$

At the interface, the energy flow of the incident wave in terms of its intensity must be conserved and is split between the transmitted wave and the reflected wave.

Hence $I_i = I_t + I_r$. Alternatively, $I_t = I_i - I_r$ (Figure 2.8b).

The intensity transmission coefficient $T_i = I_t / I_i$ and from above it can be shown that $T_i = 1 - R_i$. For example, if 0.01 (1%) of the incident intensity is reflected, then the other 0.99 (99%) must be transmitted across the boundary.

The law of reflection

In this description of reflection, it has been assumed that the interface is large compared to the wavelength of the wave and that the wave approaches the boundary at 90° (normal incidence). Under these circumstances, the reflected and transmitted waves also travel at 90° to the interface. In clinical practice, the wave may approach the interface at any angle. The angle between the direction of propagation and a line at 90° to the interface (the normal) is called the angle of incidence θ_i (which has been 0° so far) as shown in Figure 2.9a. Similarly, the angle between the direction of the reflected wave and the normal is called the angle of reflection θ_r.

For a flat, smooth interface, the angle of reflection $\theta_r = \theta_i$ the angle of incidence. This is referred to as the law of reflection. As will be seen in Chapter 5, reflection at strongly reflecting interfaces can lead to a number of image artefacts.

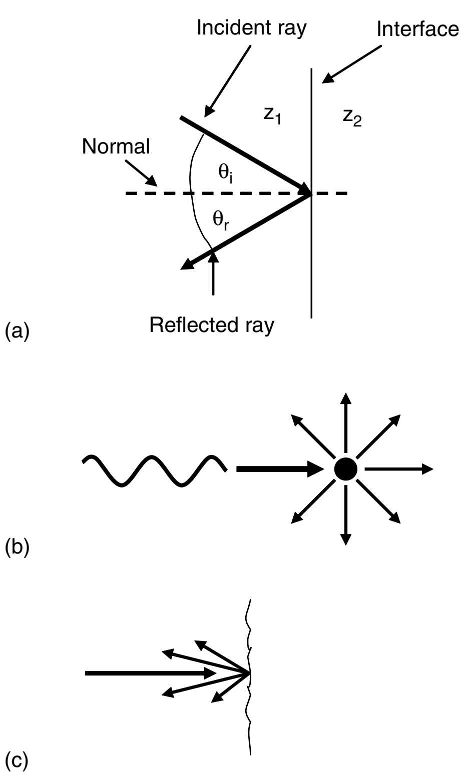

Fig. 2.9 Ultrasound waves are reflected at large interfaces and scattered by small targets: (a) at a large, smooth interface, the angle of reflection is equal to the angle of incidence; (b) small targets scatter the wave over a large angle; (c) a rough surface reflects the wave over a range of angles.

Scattering

Reflection, as just described, occurs at large interfaces such as those between organs, where there is a change in acoustic impedance. Within the parenchyma of most organs (e.g. liver and pancreas), there are many small-scale variations in acoustic properties, which constitute very small-scale reflecting targets (of size comparable to or less than the wavelength). Reflections from such very small targets do not follow the laws of reflection for large interfaces. When an ultrasound wave is incident on such a target, the wave is scattered over a large range of angles (Figure 2.9b). In fact, for a target which is much smaller that the wavelength, the wave may be scattered uniformly in all directions. For targets of the order of a wavelength in size, scattering will not be uniform in all directions but will still be over a wide angle.

The total ultrasound power scattered by a very small target is much less than that for a large interface and is related to the size d of the target and the wavelength λ of the wave. The scattered power is strongly dependent on these dimensions. For targets which are much smaller than a wavelength ($d << \lambda$), scattered power is proportional to the sixth power of the size d and inversely proportional to the fourth power of the wavelength, i.e.:

$$W_s \propto \frac{d^6}{\lambda^4} \propto d^6 f^4$$

This frequency dependence is often referred to as Rayleigh scattering.

Organs such as the liver contain non-uniformities in density and stiffness on scales ranging from the cellular level up to blood vessels, resulting in scattering characteristics which do not obey such simple rules over all frequencies used in diagnosis. The frequency dependence of scattering in real liver changes with frequency over the diagnostic range (3–10 MHz). The scattered power is proportional to f^m, where m increases with