CHAPTER 6
THE ULTRASOUND IMAGE: GENERATION AND DISPLAY

6.1 Basic principle of the ultrasonic image (see Fig. 6.1)

![Diagram of ultrasonic imaging process]

Fig 6.1 Basics of ultrasonic imaging

An ultrasonic transducer, T, sends a beam of ultrasound into the subject over a selected area of interest. At an acoustic boundary such as B within tissue, some of the ultrasound energy is reflected, either specularly or by scattering (see Chapter 3). Under favourable conditions, some of the reflected ultrasound will go back towards T. At the transducer, the returning echo will interact with the piezoelectric crystal and generate an electric signal.

This signal will be electronically processed and measured. The location of its origin at B will be determined.
Ultrasonic imaging involves the mapping of the pattern of echoes reflected from acoustic boundaries within tissues. Characteristic echo patterns are obtained for different tissues. The basic diagnostic parameters to be determined in ultrasonic imaging are:

(i) the size of an echo,
(ii) the distance of echo origin from the transducer.

The ultrasound beam is built by scanning with the beam on the subject. For each position of the ultrasound beam, a set of signals will be recorded along the beam path, corresponding to reflecting boundaries lying at different distances from the transducer. The set of signals produced along one beam path may be referred to as a "scan line". It represents single-dimensional information along the beam path. By sweeping the ultrasound beam across the subject ("scanning") in a selected direction, many other scan lines are generated to build a two-dimensional (2-D) image of a plane in the subject. This plane will be defined by the chosen direction of beam sweep. The different methods of sweeping the ultrasound beam are considered in Chapter 7.

6.2 Electronic processing of signals.

The signals generated by the returning echoes at the transducer are electronically processed and organized in computer memory before being displayed on a cathode ray oscilloscope. First, the signals are amplified to increase their sizes. The intensity of an echo may be a tiny fraction of the original output intensity of the transducer, hence the piezoelectric voltage it generates at the transducer may be very small. Secondly, echoes returning from different tissue depths must be subjected to compensation for attenuation differences. Time gain compensation (TGC) is a process of applying differential amplification to signals received from different tissue depths, with echoes originating from longer distances being amplified to a greater extent than those from shorter distances in such a way that similar tissue boundaries give equal sized signals regardless of their depth in tissue. Because the dynamic range of signal sizes may be very wide, the range of signal sizes is compressed by using logarithmic amplifiers (the compressing effect of a logarithmic function has been illustrated in the section on decibels in Chapter 4). Finally, signal sizes which are extremely small can be electronically rejected. Very
small signal sizes have an increased probability of being associated with artefacts. Rejection eliminates all signals whose magnitudes are below a certain threshold level (the rejection level).

The accepted signals are organized in computer memory before being presented to a cathode ray oscilloscope (CRO) for display. Each echo signal is associated with its own intensity level and anatomical position in tissue. Intensity and geometrical coordinates must therefore be assigned to each accepted signal. This information is then read out of memory and displayed as an image on a CRO. Hard copies of the image may then be captured using a suitable recording medium, such as thermal printing paper.

**6.3 Echo ranging**

One of the essential parameters to be determined in ultrasonic imaging is the distance between the transducer and a reflecting interface. This is done by measuring the time interval between the transmission of a pulse from the transducer and the reception of its echo back at the transducer. To achieve this, two events are electronically "marked", the moment the transducer is pulsed, and the moment it receives the returning echo from the tissue boundary. If \( d \) is the distance of a reflecting boundary (B) from the transducer (T), and \( t \) is the measured time interval between pulsing and reception, then the ultrasound beam will have travelled a distance equal to \( 2d \) (to-and-fro journey) in the time \( t \) (see Fig 6.2). To determine the distance \( d \) of the reflector, we use the simple relationship.

\[
\text{Distance} = \text{velocity} \times \text{time}
\]

or \[2d = v \cdot t\]

where \( v \) = velocity of ultrasound in the transmitting medium.
In effect, distance measurements are determined indirectly by measuring electronically the go-return time of ultrasound between the transducer and the reflecting interface. In practice, the electronic time measuring device (a cathode ray oscilloscope) is calibrated to provide distances directly.

Ultrasound systems use the average soft tissue velocity value of 1,540 m/s to calibrate distance measurements. It is interesting to consider the effect of assuming a constant value of the velocity of ultrasound in tissue in the distance calibration of ultrasound equipment, knowing that the velocity does, in fact, vary somewhat in different body tissues. How large are the errors that might be introduced in range measurement by making this assumption? The velocities of ultrasound among the soft tissues do not vary much, and most of them are quite close to their average value. Errors in distance measurement are therefore quite small. For example, the errors introduced in a range measurement of 20 cm during abdominal scanning would be of the order of 2 mm, or 1%, which is small enough to be overlooked. In contrast, the velocities of ultrasound in bone and in gas vary by large amounts from the soft tissue average, and if range measurements were to be made across any substantial distances containing those materials, large errors would be inevitable. Fortunately, this is a matter of academic interest only, since bone and gas are both acoustic barriers (see Chapter 3).

Echo ranging is possible only when ultrasound is used in the pulsed mode, in which the transducer operation is made to alternate between transmission and reception. With
continuous wave ultrasound, ranging is not possible because the time interval between pulsing and echo reception cannot be determined.

6.4 Display modes

Once the diagnostic information has been acquired and electronically processed, it has to be displayed for viewing and recording. Different methods are used to display the information acquired from ultrasound examinations. The commonly used modes are outlined in this section.

6.4.1 The amplitude mode (A-mode)

In the amplitude mode, the signals from returning echoes are displayed in the form of spikes on a cathode ray oscilloscope (CRO), traced along a time base (see Fig 6.3). On one axis (vertical axis in Fig 6.3) the amplitude of the signal (magnitude of the voltage pulse) is displayed, and on the other axis (horizontal), the position of the signal on a time scale is represented.

The amplitude of a spike is a relative measure of echo size. Because of the relationship between the distance of a reflector and the time of echo reception, the position of a spike along the time base is a measure of the distance of the associated reflecting boundary from the transducer.
The A-mode suffers from the limitation of displaying only 1-D information, representing the echoes lying along the beam path. The information does not constitute an image. Additionally, the display has the disadvantage of taking up a lot of space on the CRO in relation to the amount of information that it provides.

6.4.2 The brightness mode (B-mode)

In the brightness mode, signals from returning echoes are displayed as dots of varying intensities. The spike of the A-mode is replaced by a small dot which occupies much less space on the CRO. The intensity of a dot (the brightness) is a relative measure of echo size, with large echoes appearing as very bright dots, while at the other extreme non-reflectors appear totally dark. As in A-mode, the signals are presented along a time base on the CRO. The position of a dot along the time base is a measure of the distance of the associated reflector from the transducer. For any given position of the beam direction (scan line), a line of dots is displayed on the CRO, corresponding to the 1-D information of reflectors lying along the scan line. When the beam is swept across a selected section of the subject (the process of scanning), different dot lines are created for each scan line. These different dot lines are displayed at different positions on the CRO,
displaced laterally from one another, in relation to their corresponding beam positions. 
The combined information from different scan lines provides a 2-D image of the 
cross-section through which the beam sweeps. One dimension represents depth 
information, while the other represents lateral variations in the direction of beam sweep. 
Fig 6.4 illustrates the relationship between the positions of scan lines and the display of 
dots on the CRO to build the 2-D image.

![Diagram](image)

**Fig 6.4** Display of information in B-mode: Each dot corresponds to 
a reflecting interface along the scan line

The speed and the rate at which the ultrasound beam is swept across the subject will 
determine whether a static or a real-time image will be generated (see Chapter 7).

**6.4.3 The motion mode (M-mode)**

The motion mode is used to generate an electronic trace of a moving object lying along 
the path of the ultrasound beam. The transducer is placed in one fixed position in relation 
to the moving structure. Returning echoes are displayed in the form of dots of varying 
intensity along a time base as in B-mode. Dots for stationary reflectors will remain in the
same positions along the time base, but dots for reflectors which move in the direction of
the scan line will change their positions along the time base, because their distances from
the transducer will be changing with time. To capture the time variation of moving
structures graphically, dot lines obtained at different moments are recorded at different
lateral positions on the CRO. This is achieved by applying on the CRO an electronic
sweep of the dot lines perpendicularly to the direction of the time base (see Fig 6.5). A
time trace of the dot lines along the beam path is thus obtained. It should be noted that
the sweep of dot lines on the CRO is achieved by purely electronic means.

Fig 6.5  M-mode: Traces of dot lines for moving structure M and
stationary structure S.

Results: Stationary structures whose dots do not shift positions will trace straight
lines perpendicularly to the time base, while moving structures will trace zig-zag or
sinusoidal patterns.

The M-mode provides 1-D information along the beam path. It should be noted that
for a moving structure to be detected, it must lie along the ultrasound beam path. The M-
mode is particularly useful in examining cardiac motion.
6.4.4 Real-time mode

Real-time imaging is rapid B-mode scanning to generate images of a selected cross-section within the subject repetitively at a rate high enough to create the motion picture impression. A rapid succession of images of the same plane are generated and viewed as they are acquired. Although in reality each image in the series represents an independent static image, the effect of rapid acquisition and viewing at rates exceeding about 25 image frames per second creates the impression of continuity in time. This impression arises due to limitations in human perception. We are unable to distinguish in time between events occurring at intervals shorter than about 40 milliseconds - they appear to us to occur "at the same time".

The evolution of ultrasonic imaging into the realm of real-time was a milestone in diagnostic imaging. It is now considered a necessity in the practice of clinical ultrasound. The design of transducers capable of achieving the high framing rates required for real-time imaging is discussed in Chapter 7.

6.4.5 The Doppler mode

Before discussing the Doppler mode as a tool in clinical ultrasound, it is appropriate to introduce the Doppler phenomenon in general. The Doppler effect is observed in the behaviour of sound as well as light. In acoustics, it is associated with relative motion between the source of sound and the receiver, resulting in an apparent difference in frequency between that emitted by the source and that perceived by the receiver. An approaching sound source is perceived to be emitting sound at a higher frequency than it actually is, while a receding source appears to emit at a lower frequency. This situation arises because the wave fronts in the pressure wave of an approaching source are pushed closer together, while the wave fronts in a receding source are pulled further apart (see Fig 6.6).
Fig 6.6 Doppler effect: Wavefronts are compressed for approaching source, and decompressed for receding source, as perceived by the receiver, R.

The apparent difference in frequency is called the **Doppler shift**. For a stationary source, the wave fronts are neither compressed nor stretched, hence no shift of frequency is observed. The Doppler shift can be measured and used to:

- detect motion
- determine the direction of motion
- determine the velocity of a moving structure.

In clinical ultrasound, the Doppler mode is used in studies of blood flow and cardiac movements. When a beam of ultrasound emitted by a transducer at constant frequency interacts with a moving acoustic boundary, the boundary will, through the echoes it sends back to the transducer, act as a secondary source of ultrasound for the transducer (effectively, the reflecting boundary becomes the source, while the transducer serves as the observer). Because the boundary is moving, the transducer will detect the echoes with a Doppler shift in frequency, being of higher frequency if the interface is approaching, or of lower frequency if the interface is moving away.

Both continuous wave (CW) and pulsed wave (PW) techniques are used in Doppler ultrasound. In CW Doppler units, the transducer assembly has separate crystal elements for producing the ultrasound beam and for detecting the echoes. One crystal
continuously emits and the other continuously receives, it is not possible for one and the same crystal to transmit and detect ultrasound at the same time. By comparing the frequency of the echoes with that of the transmitted beam, it is possible to study motion (see Fig 6.7). The shift of frequency is related to the velocity of the moving reflector, and to the direction of motion. The greater the Doppler shift, the higher the velocity of the moving structure, and a higher detected frequency implies relative motion towards the transducer, while a lower detected frequency implies a receding reflector.

![Doppler shift in flow studies](image)

**Fig 6.7 Doppler shift in flow studies**

The distance of a moving structure from the transducer cannot be determined by CW ultrasound, since the go-return time for the ultrasound beam will not be known. Determination of range requires the use of pulsed beams. More sophisticated Doppler shift equipment utilizes PW ultrasound in conjunction with B-mode scanning to detect movement and determine range, and to produce images of regions of movement.

### 6.5 Ultrasound equipment and display modes

The general purpose equipment for diagnostic ultrasound will typically be a B-mode scanner capable of generating real-time images. It will be provided with 2 or 3 transducers of different frequency to cater for imaging of various organs. It will have a printer and “image freeze” capability to facilitate printing of the displayed image. The unit may have an M-mode facility incorporated for the study of motion.
Doppler ultrasound is more demanding in terms of both the equipment and the human resources required. It should be offered at specialized centres of patient care where well-trained personnel with specialized skills are available. Where the use of such equipment is appropriate, the ultrasound unit will typically combine real-time imaging with an M-mode option and Doppler facilities of varying complexity.