Resolution in ultrasound imaging

Alexander Ng MB ChB FRCA MD
Justiaan Swanevelder MB ChB FRCA FCA(SA) MMed

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Key points

Spatial resolution of images is enhanced by short spatial pulse length and focusing.

Compared with low-frequency pulses, high-frequency pulses have shallow depth of penetration owing to increased attenuation.

Temporal resolution of a two-dimensional image is improved when frame rate is high.

When an image is displayed in one dimension over time, temporal resolution is high.

Contrast resolution is altered by compression of the range of reflected ultrasound amplitudes, number of layers of bits per pixel, and the use of contrast agents.

Alexander Ng MB ChB FRCA MD
Consultant Anaesthetist
Heart and Lung Centre
Royal Wolverhampton Hospitals NHS Trust and University of Birmingham
West Midlands WV10 0QP
UK
Tel: +44 1902 307999 ext. 4357
Fax: +44 1902 694352
E-mail: alexander.ng@rwh-tr.nhs.uk
(for correspondence)

Justiaan Swanevelder MB ChB FRCA FCA(SA) MMed
Consultant Anaesthetist
Glenfield Hospital
University Hospitals of Leicester NHS Trust
Leicester LE3 9QP
UK

Ultrasound scanning is now utilized in all aspects of anaesthesia, critical care, and pain management. Typical applications include determination of left ventricular function and cardiac output, assessment of haemodynamic instability, assistance with difficult venous access, and facilitation of accurate neural block.[1–3] One aspect of competency in ultrasound imaging includes an understanding of how images can be displayed optimally.[4] This article discusses three main aspects of the physics of diagnostic ultrasound, that is to say, spatial resolution, temporal resolution, and contrast resolution; it utilizes examples from peri-operative echocardiography to illustrate these principles.

Spatial resolution

To understand how an image on the screen of an ultrasound system is produced, it is necessary to examine the features of a transducer and the ultrasound beams that it creates and receives. A transducer consists of many piezoelectric elements that convert electrical energy into sound energy and vice versa.[5] Ultrasound, in the form of a pulsed beam, propagates from the surface of the transducer into soft tissue. Sound waves are absorbed in part by tissue but are also reflected back to the transducer where they are detected. Ultrasound scanners are able to process many pulsed beams instantly and thus create real-time images for diagnostic use. The ability of an ultrasound system to distinguish between two points at a particular depth in tissue, that is to say, axial resolution and lateral resolution, is determined predominantly by the transducer.

Axial resolution

Axial (also called longitudinal) resolution is the minimum distance that can be differentiated between two reflectors located parallel to the direction of ultrasound beam. Mathematically, it is equal to half the spatial pulse length. Axial resolution is high when the spatial pulse length is short.

Spatial pulse length is the product of the number of cycles in a pulse of ultrasound and the wavelength (Fig. 1A). Most pulses consist of two or three cycles, the number of which is determined by damping of piezoelectric elements after excitation: high damping reduces the number of cycles in a pulse and hence shortens spatial pulse length (Fig. 1B). The wavelength of a pulse is determined by the operating frequency of the transducer; transducers of high frequency have thin piezoelectric elements that generate pulses of short wavelength (Fig. 1B). The wavelength is equal to twice the thickness of the elements in the transducer.

These potentially desirable characteristics, that is to say, damping and high frequency, have the following problems related to attenuation.

(i) Excessive damping is associated with loss of amplitude and hence low-intensity ultrasound (Fig. 1B).
(ii) High-frequency pulses are attenuated well in soft tissue which means that they may not be reflected back sufficiently from deep structures, for detection by the transducer.

Attenuation is expressed in decibels and is determined by both the frequency of ultrasound and depth of the reflector from the transducer. Assuming an attenuation coefficient in soft tissue of 0.5 dB cm^{-1} travelled by ultrasound of 1 MHz, we find that:

$$\begin{aligned}&\text{Attenuation in decibels}\\&\quad = 0.5\times 2\times \text{depth of reflector in cm}\\&\qquad \times \text{frequency in MHz}\end{aligned}$$

where

$$\begin{aligned}&\text{Attenuation in decibels}\\&\quad = 10\times \log_{10}\left(\frac{\text{received intensity}}{\text{propagated intensity}}\right)\end{aligned}$$

At a particular frequency, increasing attenuation at longer depths from the transducer is minimized by progressive amplification of the

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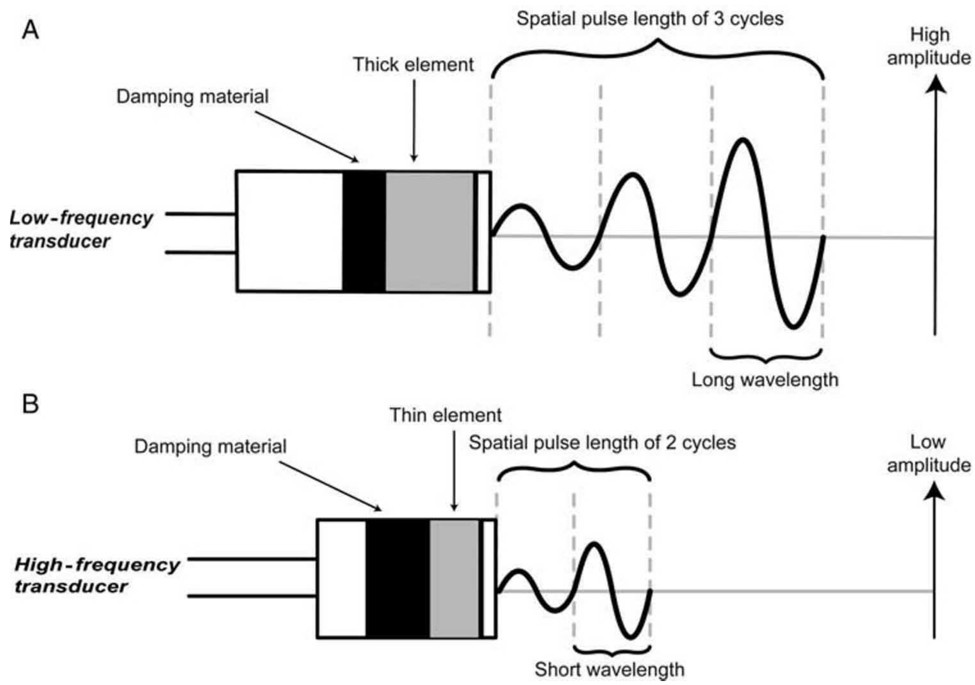

Fig 1 (A) Low-frequency transducer with long spatial pulse length and low axial resolution. (B) High-frequency transducer with short pulse length and high axial resolution.

power of the reflected pulses. This process is called 'Time Gain Compensation' and is expressed as a logarithmic ratio of power, in decibels, similar to attenuation.

$$\text{Gain in decibels} = 10 \times \log_{10}\left(\frac{\text{output power}}{\text{input power}}\right)$$

Axial resolution of deep structures is comparatively low since a transducer of low frequency (2 MHz) has to be used, for example, during transthoracic imaging of the heart. In this situation, excessive attenuation would preclude the use of high-frequency transducers, even after amplification of reflected ultrasound by Time Gain Compensation. The alternative consideration is to locate the transducer close to the structure of interest so that a high-frequency transducer can be used with high axial resolution. Examples of this practice in anaesthesia include transoesophageal (7 MHz) and epiaortic (10 MHz) placements of the transducer.[6,7]

Lateral resolution

Lateral resolution, with respect to an image containing pulses of ultrasound scanned across a plane of tissue, is the minimum distance that can be distinguished between two reflectors located perpendicular to the direction of the ultrasound beam. Lateral resolution is high when the width of the beam of ultrasound is narrow.

The width of the beam and hence lateral resolution varies with distance from the transducer, that is to say:

(i) At the transducer, beam width is approximately equal to the width of the transducer.
(ii) Then, the beam converges to its narrowest width which is half the width of the transducer, at a perpendicular distance from the transducer called the near-zone length (Fig. 2A). The region of space subtended by the beam is called the near zone (Fresnel's zone).
(iii) At a distance greater than the near-zone length, that is to say in the far zone (Fraunhofer's zone), the beam diverges such that it becomes the width of the transducer, when the distance from the transducer to the reflector is twice the near-zone length. Here, lateral resolution decreases.

Lateral resolution is high when near-zone length is long. Near-zone length is determined by factors contained in the equation:

$$\text{Near zone length} = \frac{\text{diameter}^2}{4 \times \text{wavelength}}$$

Factors that increase near-zone length include:

(i) short wavelength;
(ii) high frequency of transducer, comprising thin piezoelectric elements with high damping (frequency and wavelength are inversely related);
(iii) large aperture (wide element width).

When low-frequency transducers are utilized, lateral resolution deteriorates: in this situation, near-zone length is short which

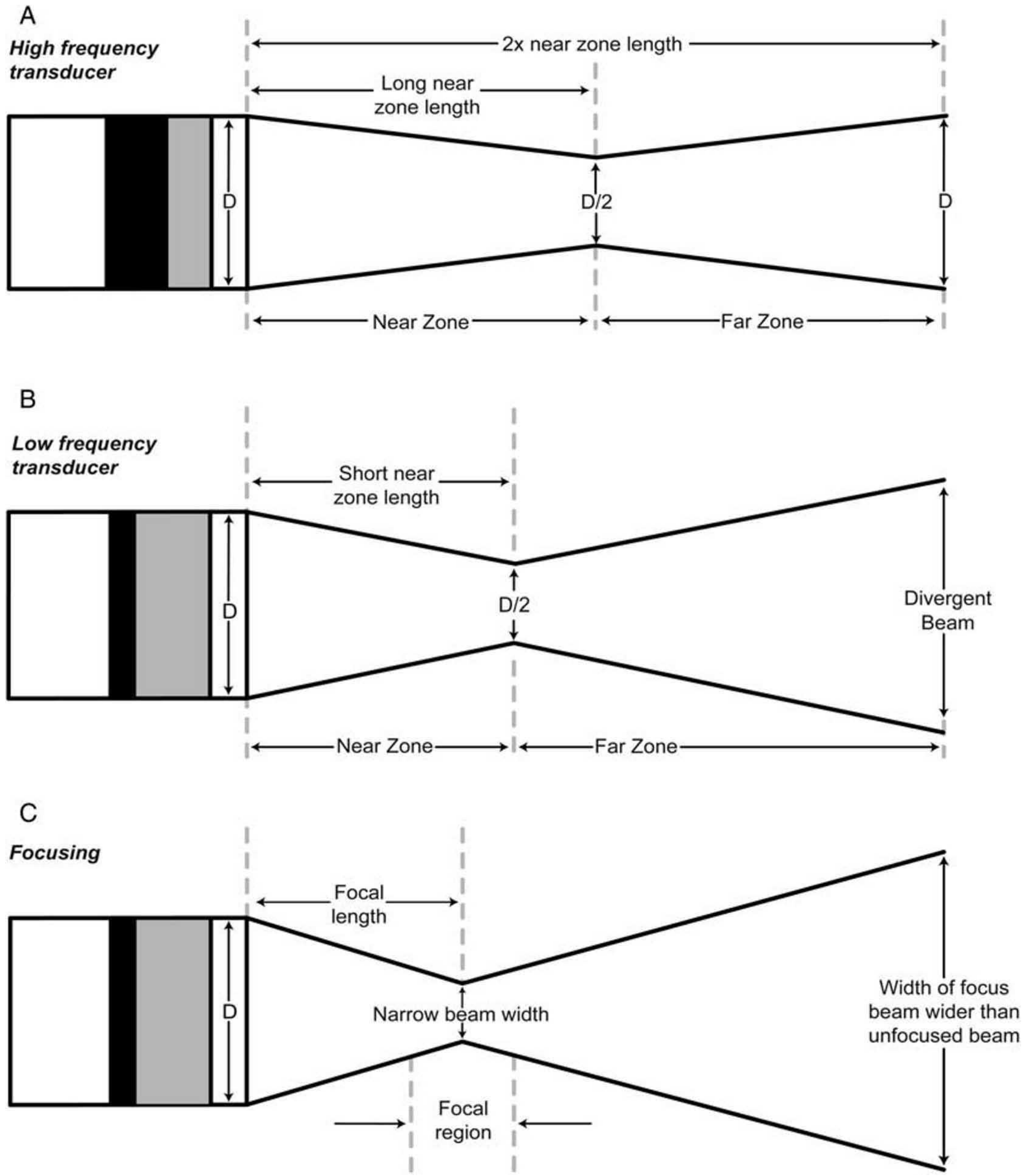

Fig 2 (A) High-frequency transducer with long near-zone length and narrow beam width. (B) Low-frequency transducer with short near-zone length and wide beam width. (C) Focusing narrows beam width.

means that structures beyond the near zone are scanned by a divergent beam (Fig. 2B).

In addition, extraneous beams (called grating lobes) surrounding the main beam from a multi-element transducer may cause artifact and reduce lateral resolution. Grating lobes may be minimized by driving the elements at variable voltages in a process called apodization.

Focusing

Piezoelectric elements in a transducer operate at different times and can narrow the pulsed beam with improved lateral resolution. This process of focusing leads to the creation of a focal region within the near zone, but not the far zone (Fig. 2C). Focusing shortens the distance of the narrowest point of the beam from the transducer, that is to say, it reduces the near-zone length to a shorter value called the focal length. The beam beyond the focal region is divergent and so there is a reduction in lateral resolution of structures deeper than this point. The width of a focused beam is determined by factors contained in the following equation:

$$\text{Width of focused beam} \approx \frac{\text{focal length} \times \text{wavelength}}{\text{aperture}}$$

Thus, a narrow, focused beam, and hence high lateral resolution, is obtained by:

(i) short focal length;
(ii) short wavelength;
(iii) wide aperture.

In addition, it is possible to improve lateral resolution by focusing at more than one depth within tissue. This process requires repetition of pulses of ultrasound along the same scan line for each focal point. Unfortunately, there is a concomitant reduction in frame rate and hence temporal resolution. Although this practice of multiple foci may be suitable for imaging sessile structures, for example, nerves, it is less advisable for rapidly moving structures such as the heart for which one focus would be optimal.

Temporal resolution

Anatomical structures are displayed on the screen of the ultrasound machine, in two or three dimensions, as sequential frames over time. Each frame is created from repeated pulses that form scan lines; these may be duplicated depending on the number of focal points (Fig. 3A). Temporal resolution is the time from the beginning of one frame to the next; it represents the ability of the ultrasound system to distinguish between instantaneous events of rapidly moving structures, for example, during the cardiac cycle. A high frame rate and hence enhanced temporal resolution may be improved by:

(i) reduced depth of penetration, since pulses have to travel a short distance;
(ii) reduced number of focal points, since scan lines do not have to be duplicated;
(iii) reduced scan lines per frame, using narrow frames rather than wide frames.

This relationship may be derived from the following equation:

$$\text{Distance} \times \text{frequency} = \text{speed}$$

where

$$\text{Distance of propagation and reflection} = 2 \times \text{depth of penetration of ultrasound}$$

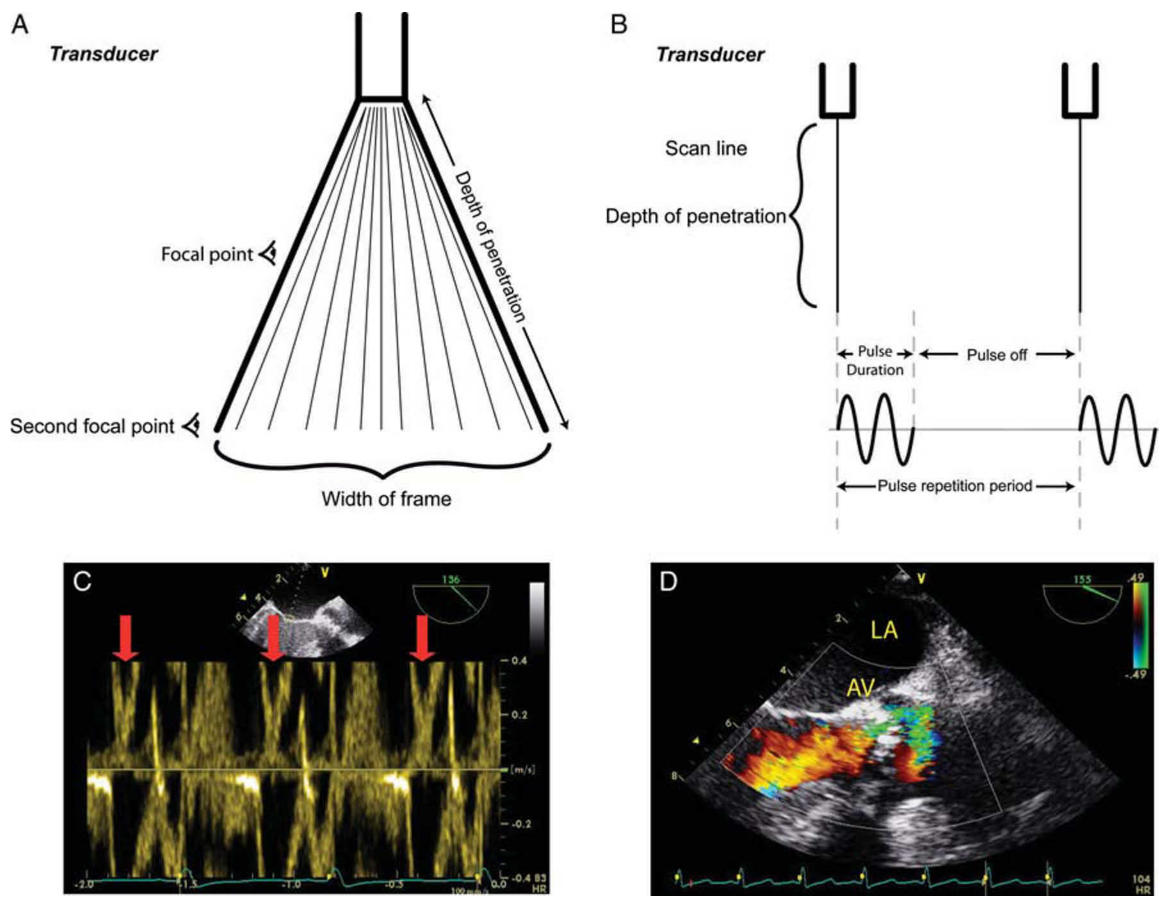

Fig 3 (A) A frame comprising many scan lines displays structures in two dimensions. (B) In M mode displaying depth over time, the scan lines are transmitted at the pulse repetition frequency. (C) Pulsed-wave spectral Doppler showing aliasing of the mitral E-wave (red arrows). (D) Colour Doppler imaging of the left ventricular outflow tract, calcific aortic valve (AV) with stenosis. Red colour represents blood flow towards the transducer. Flow accelerates through the AV (shown in green). LA, left atrium.

Frequency, that is to say, pulse repetition frequency

$$= \text{frame rate} \times \text{number of foci} \times \text{number of scan lines per frame}$$

Speed of propagation of sound in tissue

$$= 1540\ \text{m s}^{-1} \text{ or } 154\,000\ \text{cm s}^{-1}$$

Putting the values in and rearranging the equation, we can obtain the frame rate and hence temporal resolution.

Frame rate

$$= \frac{154\,000}{2 \times \text{penetration in cm} \times \text{number of foci} \times \text{number of scan lines}}$$

In contrast to the above display of anatomical structures within repeated frames derived from many scan lines, it is possible to obtain improved temporal resolution when using motion (M) mode. In M mode, structures are represented in one anatomical dimension as each scan line represents new information over time. Temporal resolution is defined by the time taken from one pulse to the next, that is to say, the pulse repetition period (Fig. 3B). High temporal resolution is obtained when the pulse repetition period is short; this is achieved by a high pulse repetition frequency when the depth of penetration is shallow. In clinical practice, extremely good temporal resolution may be attained with M mode, for example, when assessing the onset and rate of wall thickening of opposite segments of the myocardium and when assessing movement of valvular structures.

Doppler

In Doppler mode, pulses of ultrasound travel from a transducer to a moving target where they are reflected back towards the transducer. This process is intermittent and occurs at a frequency called the pulse repetition frequency. In contrast to imaging mode, the spatial pulse length is long since each pulse contains 5–30 cycles.

The frequencies of the waveforms of received and transmitted pulses are analysed and the difference between them is called the Doppler shift frequency. Doppler shift frequency is useful primarily because it enables the velocity of the reflector (e.g. red cells in blood) to be measured, as shown in the Doppler equation.[8]

Doppler shift frequency

$$= \frac{2 \times \text{operating frequency} \times \text{velocity of reflector} \times \cos\theta}{\text{speed of sound}}$$

Doppler shift frequency obtained from the difference between received and propagated pulses varies. If it reaches a value that is half of the frequency of the intermittently propagated pulses (pulse repetition frequency), temporal resolution of the Doppler image will be affected with incorrect representation of velocity. This Doppler shift frequency is called the Nyquist limit.

$$\text{Nyquist limit} = \frac{\text{pulse repetition frequency}}{2}$$

A pulsed spectral Doppler display of Doppler shift frequency (and hence velocity) with time has a low maximum Doppler shift value (Nyquist's limit) beyond which temporal aliasing occurs (Fig. 3C). There are three main ways to improve temporal resolution and hence prevent temporal aliasing.

(i) As derived from the Doppler equation, a transducer operating at a reduced frequency can be used to keep the Doppler shift value less than the Nyquist limit for the same velocity of reflector.
(ii) Alternatively, pulses can be sent at a high pulse repetition frequency, with some loss of depth resolution, called range ambiguity.
(iii) Finally, pulses can be sent at the transducer's high fundamental frequency (continuous wave spectral Doppler mode rather pulsed spectral Doppler mode) so that very high Doppler shifts and hence very high velocities can be measured. However, depth resolution is no longer possible with this modality.

Furthermore, Doppler information may also be displayed in a colour window (colour Doppler mode) similar to the frames for imaging anatomic structures described above (Fig. 3D). Each frame of the colour window is formed from repeated pulses of ultrasound. A frame comprises scan lines, each of which has multiple pulses (called ensemble length). These repeated pulses per scan line are needed to provide sufficient information for autocorrelation, a process in which information contained in the Doppler shift frequencies is used to determine the sign (direction of movement of reflector), mean velocity of reflector, power, and variance (the spread of velocities). Frame rate is extremely low and is influenced by the factors contained in the equation below.

Frame rate

$$= \frac{154\,000}{2 \times \text{penetration in cm} \times \text{ensemble length} \times \text{number of scan lines}}$$

Two aspects concerning temporal resolution are pertinent to colour Doppler imaging:

(i) Frame rate and hence temporal resolution may be improved by utilizing narrow colour windows.
(ii) The maximum magnitude of the velocity detected by colour Doppler may be altered by the ultrasonographer; by doing so, there is a concomitant alteration in the frequency of propagated pulses (pulse repetition frequency). Lowering of the magnitude of velocity and the transducer's pulse repetition frequency leads to deliberate reduction in temporal resolution, so that aliasing occurs for the detection of low velocities or for

specific measurements, for example, regurgitant orifice area by the proximal isovelocity surface area method.

Contrast resolution

Contrast resolution refers to the ability to distinguish between different echo amplitudes of adjacent structures. Contrast resolution may be enhanced at various stages in the imaging process, these include compression, image memory, and the use of contrast agents.

Compression

Pulses of ultrasound vary in amplitude and hence power. The magnitude of the highest to the lowest power is expressed logarithmically, in a decibel range called dynamic range.

$$\text{Dynamic range in decibels} = 10 \times \log_{10}\left(\frac{\text{highest power}}{\text{lowest power}}\right)$$

Compression occurs in the signal processor which reduces the dynamic range, that is to say, the ratio of the highest power to the lowest power, by a combination of assigning stronger echo powers to maximum and weaker echo powers to zero. Numerically, the number of decibels of the dynamic range is reduced by compression. High compression with a narrow dynamic range (e.g. 30 decibels) creates an image of high contrast (Fig. 4A). Conversely, low compression with wide dynamic range (e.g. 60 decibels) displays an image of low contrast and with many shades of grey (Fig. 4B). Contrast resolution may be enhanced by adding colour since the human eye can distinguish between more shades of colour than shades of grey (Fig. 4C).

Image memory

Storage of digitized information contained in the pulse waveforms occurs in the image memory. To enable various shades of grey to be visualized, each part of the image memory called a pixel (picture element) must have as many layers of bits (binary digits) as possible. Each bit contains a code of 0 or 1. Taking an example of a pixel which has five layers, we find that the number of shades of grey is derived from the sum of the maximum numbers for the binary digits in each layer, shown as:

2^0	2^1	2^2	2^3	2^4	Total
1	2	4	8	16	31

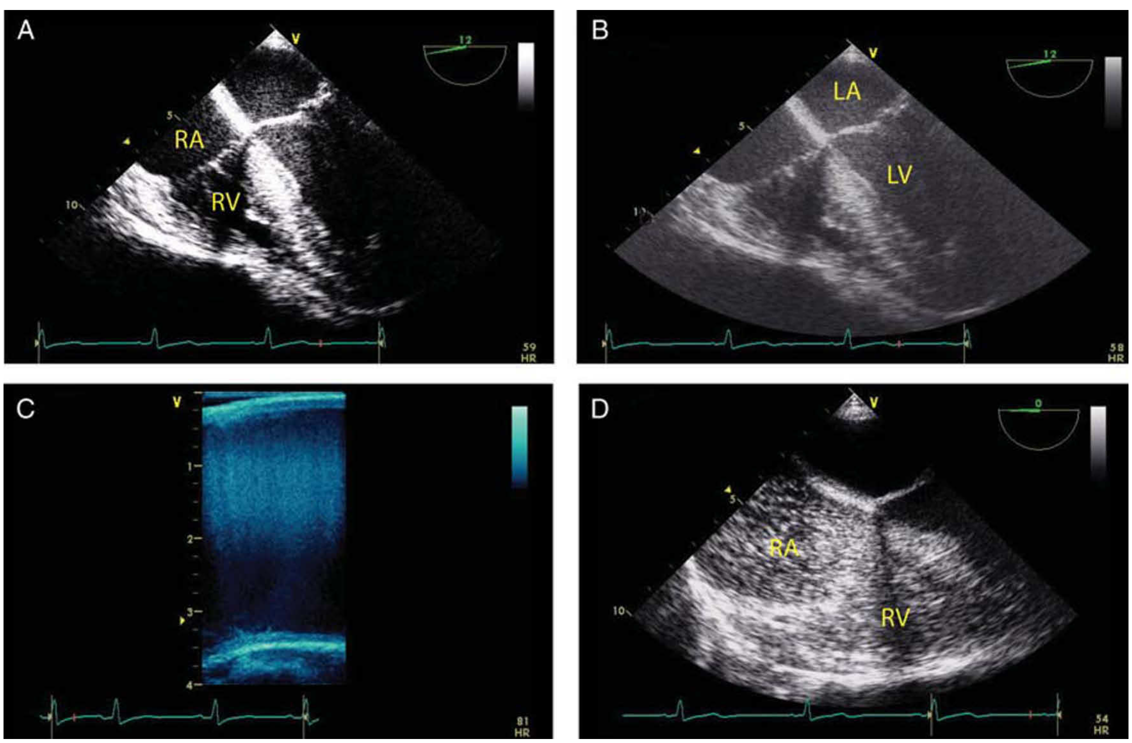

Fig 4 (A) Mid-oesophageal transoesophageal echocardiographic image of the left ventricle (LV), right ventricle (RV), left atrium (LA), and right atrium (RA). The image is of high contrast owing to high compression and a narrow dynamic range. (B) Mid-oesophageal transoesophageal echocardiographic image of the LV, RV, LA, and RA. This image is of low contrast owing to low compression and wide dynamic range. (C) Aqua colour to improve contrast of the proximal ascending aorta obtained by epiaortic imaging during cardiac surgery. (D) Mid-oesophageal transoesophageal echocardiographic view of the RA and RV showing bubbles of agitated saline. These bubbles reside in the right heart and their appearance contrast with their absence in the left heart.

The total of the numbers including 0 is 32 and thus a 5 bit memory enables 32 shades of contrast to be stored.

Contrast agents

Contrast agents are used when conventional ultrasound imaging does not provide sufficient distinction between myocardial tissue and blood. Contrast agents are suspensions of microbubbles of gas, for example, agitated saline, perfluoropropane or sulphur hexafluoride.[9] After administration, they reside temporarily in blood and may be visualized separately from the myocardium. This phenomenon arises because the impedance for ultrasound in gas is markedly different from that for soft tissue. Impedance is the product of density and propagation speed, and it can be appreciated that impedance in air is low whereas that in soft tissue is high. When such a disparity occurs, ultrasound is reflected strongly from the microbubbles, thus enhancing contrast resolution and visualization of structures of interest (Fig. 4D).

However, strong reflection and high contrast are not always desirable. They occur naturally when a transducer is placed on the tissue of interest where two main boundaries of different impedances are created. The first boundary occurs between the element of a transducer and air, whereas the second boundary occurs between air and the tissue of interest. To obviate strong reflection and hence promote transmission of ultrasound, a medium of intermediate impedance has to be present between the two sides of the boundary. For the element–air boundary, there are matching layers on the surface of the transducer, and for the air–tissue boundary, a coupling medium (gel) is applied. In this way, adverse contrast is minimized.

Conclusion

In conclusion, resolution of ultrasound information is affected by several factors considered above. A thorough understanding of these factors will enhance both quality and interpretation of data contained in the images. By doing so, the ultrasonographer provides useful information for clinical decisions and hence may contribute to improved outcomes in the perioperative period.[10]

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Conflict of interest

A.N. is a member of the editorial board of *CEACCP*.

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Please see multiple choice questions 29–32.