



EFSUMB Course Book, 2nd Edition

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Ultrasound “Knobology”

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INTRODUCTION

Ultrasound is fast becoming the clinicians' stethoscope and whilst the number and range of controls available on a scanner can initially look daunting, all scanners tend to have the same basic image controls. Although these controls may be given different names, they tend to have similar image icons and be located on similar positions on the main control panel to optimise user ergonomics. This section provides a guide on the functionality of ultrasound scanner controls and how to utilise them in routine clinical scanning. The aim is for the recently initiated to understand the basic scanner controls in order to optimise the ultrasound image and obtain useful diagnostic information.

GETTING STARTED

On/Off switch

There is usually an "On/Off" control button on the keyboard console. On some scanners however, it is to be found on the panel near the port for the transducers and occasionally on the back of the scanner where all the output cables feed into. Upon the start-up, the configuration of the scanner is usually complete within 30 seconds but on older models this can take substantially longer. After a scanning session is complete or at the end of the day, it is usually advisable to power down the scanner before turning it off at the mains.

Patient details

Entering the patient details is the next key step after configuration of the scanner is complete and this is usually denoted by a 'face' icon or 'ID' in letters. When connected to the hospital network, the user should select 'work list' and highlight the correct patient from the work list.

TRANSDUCER CHOICE:

Type of transducer

Dependent on the scanning department, there may be a range of transducers available to use – the most frequently used ones are likely to be hanging on the side of the scanner in the holders. Commonly used transducers include linear and curvilinear arrays. Linear arrays tend to be higher frequency and hence produce higher resolution images than curvilinear arrays but have lower penetration. They are optimally used for imaging superficial or small objects in detail and are used routinely for imaging breast, cerebrovascular, musculoskeletal and peripheral vascular structures producing a rectangular image field of view with uniform image line density. Curvilinear arrays, which as the name suggests have a curved surface, tend to operate at lower frequencies compared with linear arrays and are used for imaging deep abdominal structures (e.g. gynaecology and obstetrics). The curvilinear transducer produces a fan-shaped image and is thus capable of imaging without the surface of the transducer. However, line density decreases at increasing depth, which will affect the resolution of the image at depth. Phased array transducers are used for imaging at depth in areas of restricted access as they have a small footprint (e.g. cardiology where the ribs limit the scanning window). They also produce a fan-shaped image.

Frequency of transducer

As the ultrasound wave propagates through soft tissue, the intensity of the ultrasound decreases (attenuates) as a function of both depth and frequency. For tissues the total attenuation, measured in decibels (dB), is assumed to vary linearly with frequency and depth so that a doubling of the frequency (or depth) will increase the attenuation by a factor of 2. Hence, choice of a higher frequency transducer will result in greater attenuation of the ultrasound beam.

The trade-off is that higher frequencies give better spatial resolution. Spatial resolution can be defined as the ability of the scanner to differentiate between two adjacent objects and clinically relates to the smallest size of structure the user would be able to confidently differentiate from surrounding structures. In general, the higher the frequency, the better the lateral resolution and the lower the penetration depth. For example, for a 3 MHz

transducer, resolution will be of the order of 0.5-1 mm and with a depth of penetration of around 20 cm, while at 10 MHz, resolution would be 3 times better and penetration limited to less than 6 cm.

When starting a scan, it is usually better to start with the transducer that will give the best penetration (i.e. with a lower frequency) and then move to a higher frequency if improved resolution is needed to image a more superficial structure. For example, for an abdominal scan, typically select the curvilinear probe with a frequency range of between 1 and 5 MHz whilst if scanning a superficial structure, such as a lump on the skin, then select the linear 12-18 MHz probe. For vascular structures, there are optimised probes on certain scanners, which are typically in the mid-range (i.e. 6-12 MHz). The footprints of these probes will also vary and it is best to tailor this to the site, access and structure being assessed.

ESSENTIAL BUTTONS/KNOBS FOR IMAGING:

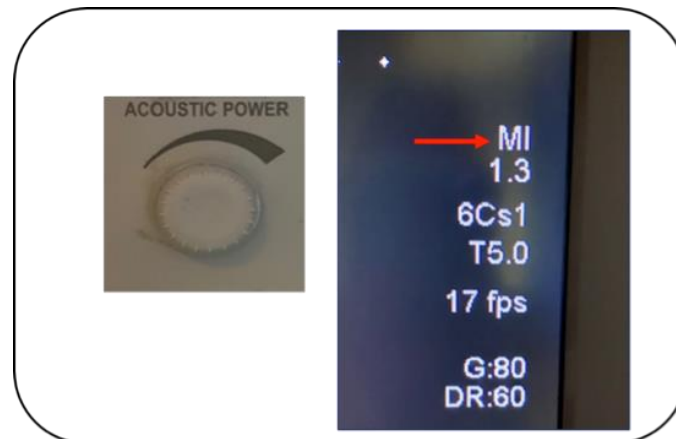
Presets

All manufacturers will have optimised preset/start-up configurations so that most settings are pre-selected and initialised for the user. This is particularly useful for new users of ultrasound. Generally, these preset configurations include initialisation of the transducer, power output (usually set to maximum), depth, focal position, dynamic range, degree of smoothing, persistence and many other controls. While the majority of these controls can be adjusted by the user individually, the presets provide a good starting set-up for each scan. Although personalised set-ups can also be saved onto the scanner, it must be remembered that many of the set-up controls are interdependent so that changing one set-up parameter is likely to cause others to be modified.

Power Output

Depending on the manufacturer, the power output button may either be located on the main control panel or be selected as an option in a set-up menu. All commercial ultrasound scanners enable the user to vary the power output of the transducer from minimal values up to a maximum value. By altering the power output, the thermal and mechanical indices (safety output indices) which are displayed on all scanner screens [Figure 1] will vary.

Figure 1 The *Acoustic Power* button is usually clearly marked and an indication of the power output is given by the mechanical index or abbreviated to MI (arrow) which is displayed on the screen.



The magnitude of the maximum power output value is related to the ultrasound-imaging mode selected (e.g. Doppler imaging versus 2D imaging) and the organ being scanned (foetus versus abdominal soft tissues). Most scanners will default to the maximum permissible power output (maximum intensity) on power-up as this will maximise the depth, which can be imaged using the selected transducer. However, users should be aware of the interplay of power output and gain to achieve an optimal diagnostic image, as similar imaging results can often be obtained by lowering the power output and increasing the gain.

Depth/Zoom/Width

Sufficient penetration (depth) and magnification of the field of view are essential to achieve an overview of the anatomy. On most scanners, the 'depth' and 'zoom' controls are clearly marked and feature as control buttons or levers, which can be incrementally adjusted to increase or decrease depth and size. The 'zoom' feature is generally located close to the depth knob and typically has a symbol similar to a magnifying glass [Figure 2]. When the 'zoom' is selected a region of interest (ROI) box appears on the image, the position and size of which can be modified using the trackball. Within the zoom window the image line

density is increased enabling enhanced anatomical detail to be visualized. In addition, higher frame rates can be obtained as only the depth of the zoom window is scanned.

Figure 2 Typical Depth/Zoom buttons and icons (arrow).



Frequency choice

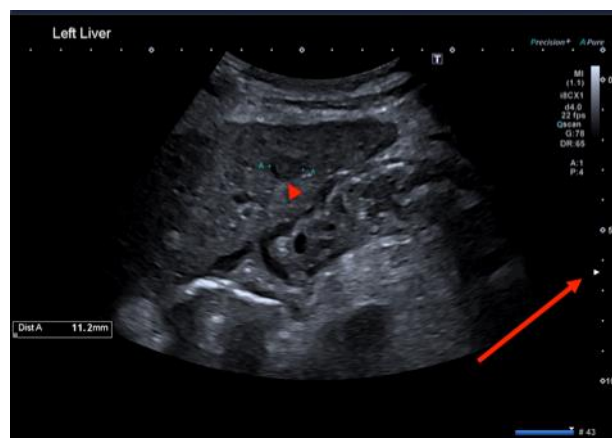
All transducers have a specified center frequency and a frequency range (bandwidth) over which the transducer will operate efficiently. For many this will be reflected in the transducer name – e.g. 'C1-5' indicating a curvilinear probe operating between 1 to 5 MHz. High-end scanners provide the opportunity to adjust the insonation frequency over this narrow frequency range using either a knob or lever switch. For some scanners rather than stating the shift in frequency, the terminology used is 'resolution', 'general' and 'penetration' with 'resolution' mode indicating lower penetration and higher resolution (higher frequency), 'penetration' indicating increased penetration with lower resolution (lower frequency) and 'general' operating at the center frequency of the transducer.

Focal Position

Optimal lateral resolution and anatomical detail are obtained within the focal zone of a transducer. Selection of a focal depth by the user will vary the time delays on the array elements within the transducer so that the ultrasound beam is narrowest at the focal depth selected. Visually, the grain structure within the image (speckle) will be finer with increasing grain size at increasing distance from the focal position.

On a scanner, this knob or lever is usually labelled as 'focus' and is located near to the 'depth' and 'zoom' buttons. The position of the focus is indicated by an arrow on the side of the image or a line indicating a range if 'range focus' is employed. The user can usually select multiple foci but the trade-off is a significantly reduced frame rate, sometimes for only a small increment in image resolution [Figure 3].

Figure 3 Focal Zone positioning (arrow) and calliper measurement of a HCC in a cirrhotic liver (arrow head).



2D Gain and Time compensation (TGC)

The '2D gain' is usually marked as such and typically controlled with a knob, which can be turned to adjust the overall brightness of the image. The degree of brightness will depend on the preference of the user and the degree of darkness of the room setting. Unlike the power output, adjustment of the 2D gain has no effect on the emitted ultrasound intensity, and in

many instances the output power can be reduced and compensated by an increase in the overall gain.

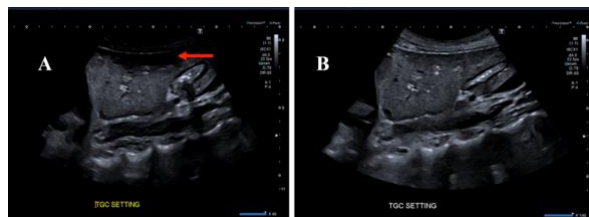
As discussed above, the ultrasound beam is attenuated as it travels through the body, a result of scattering and absorption of the ultrasound beam. Since the amount of attenuation is dependent on the frequency of the beam and also on the depth of tissue through which the beam has to travel, without any incremental adjustment the ultrasound image would be brighter nearer to the transducer and darker at depth. Time gain compensation (TGC) is used to provide a uniformly bright image for the user so that any abnormality is more easily appreciated. 'Time gain compensation' controls (5-9 individual slider controls) tend to be located on one side of the control panel of the scanner. The slider-controls correspond to different depth segments of the image and manual adjustment of these can enhance the gain at specific depths of the image [Figure 4].

Figure 4 TGC (arrow) and auto TGC (arrow head) buttons.



State-of-the-art scanners now employ an automated TGC and many also work in the background, optimising the quality and brightness of the image, often without the user realising that this is occurring during real-time scanning [Figure 5]. Manufacturers have acronyms for their auto TGC, e.g. “QScan” (Canon Medical Systems), “iScan” (Philips Medical Systems), “TEQ” (Siemens Healthineers), etc.

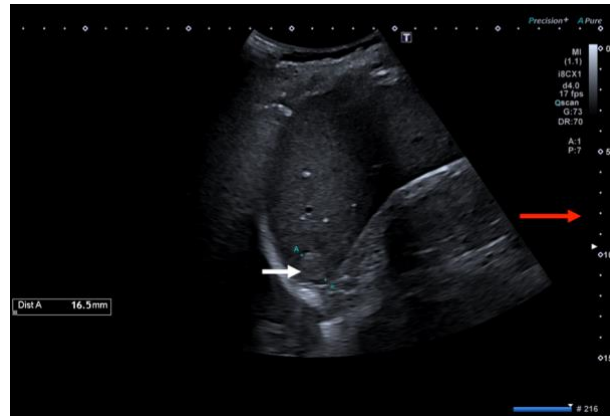
Figure 5 No TGC adjustment – note the near field of the liver is darker compared with the deeper segment (arrow, A). After TGC adjustment (B).



Measurement

The measurement tool is one of the most frequently used tools on the scanner and is used to measure the size of structures and delineate their extent. The measurement button is usually a symbol of either a ruler or callipers joined by dots. There is also a scale on the side of the image, which is in 0.5 cm or 1 cm increments depending on the size/depth of the image [Figure 6].

Figure 6 Measurement callipers outlining a liver haemangioma (white arrow). Note the scale in cm on the right-hand side of the image (red arrow).



In order to calculate the depth of a structure, the scanner assumes that the ultrasound velocity is a constant and equal to 1540 m/s. For calculation of the area of structures, usually a circle or elliptical structure is assumed and the user indicates with the callipers the start and end point of a typical diameter (circle) or major axis (ellipse). Using the trackball, the minor axis of the ellipse is then indicated and the area is illustrated on the screen. Similar processes can be undertaken to calculate volume measurements based on images acquired in orthogonal imaging planes.

Trackball/Freeze/Cineloop

The trackball or touchpad is the “mouse” of the ultrasound scanner and is the common operating device of the screen cursor. The ball can be rotated freely in all axis directions and typically many functions are controlled with it, such as scrolling through a video, positioning the body marker or positioning measurement callipers. Most scanners also have a ‘select’ or unlabelled push button adjacent to the trackball, which is similar to the left- and right-hand “click” functions found with a computer mouse [Figure 7].

Figure 7 Example of layout of function keys/symbols for ease of use – a. trackball, b. unlabelled 'select' keys, c. body marker, d. text annotation, e. measurement callipers, f. storage.



The 'freeze' function is used to pause the moving live image and there is also an automatic 'cineloop' function between two 'freeze' button clicks. This enables the user to scrutinize individual frames acquired previously more precisely. This is particularly advantageous for locating structures that were only briefly visible in the moving image and can elude targeted freezing attempts.

The length of the cineloop depends on the system used and can usually be modified by the user, e.g. longer loops associated with ultrasound studies where contrast agent is injected. The cineloop can be stored retrospectively (i.e. between two freeze frames) or prospectively starting once the 'video' function has been activated. The length of these clips varies depending on the manufacturer but can usually be altered to suit user preference.

Image/clip store

This is an important function to indicate the body part that is scanned and also to delineate any relevant abnormality. An acquired picture is treated as a legal medical document and such images are typically stored for at least 7 years depending on the laws of the respective country. The button usually features a camera or the word 'store' or 'print' and in case of a video clip/cineloop, a picture illustrating a 'reel of film' or the word 'cine' [Figure 8].

Premium scanners now store the image as raw data and therefore certain functionalities, such as measurements or altering the image brightness, become available to adjust and alter on retrieval of the image.

Figure 8 Example of 'still store' and 'cine store' buttons (arrows).

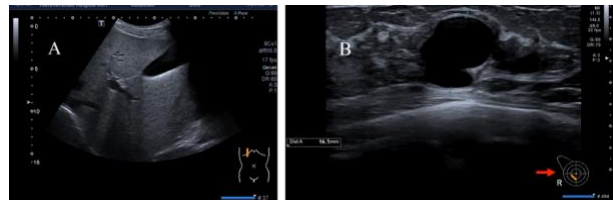


Annotation/body marking

It is important that the body part that is being scanned is indicated on the image. The ability to annotate is usually performed by pressing the 'annotate/text' button or on occasions, by an 'ABC' icon. Alternatively, some manufacturers have specific function keys assigned to this

or text will appear as soon as the keyboard is activated. At times it may be easier to indicate the position of the probe on the body part diagram. Each manufacturer will have a body part figure and the user can just select the appropriate body part and move the position of the probe using the trackball [Figure 9].

Figure 9 Liver and gallbladder image with body mark showing the position of the probe in orange (A). Example of a body mark and position of the probe indicating where the simple cyst lies within the right breast (arrow). Note the calliper markers measuring the size of the cyst (B).



ADVANCED IMAGING BUTTONS

To attain the optimal image quality, manufacturers typically suggest selecting the uploaded presets for the determined body part. However, there may be occasions where it will be useful to know the functionality of some of these “advanced” buttons in order to be able to adjust the image and/or resolution if necessary.

Harmonic Imaging

The harmonics button is usually denoted by ‘Tissue Harmonics’ or ‘Tissue Harmonic Imaging (THI)’ and is usually either an ‘on’ or ‘off’ button. Whilst the image resolution improves in harmonic mode, there is a trade-off with the depth and penetration of the image.

Harmonic imaging is used to reduce the noise in an image. When low amplitude ultrasound is transmitted into the body, the ultrasound wave is scattered in a linear manner, i.e. the scattered waves are of the same phase as the transmitted ultrasound. When the acoustic pressure is increased, non-linear propagation causes the transmitted wave to become

increasingly distorted at depth generating harmonics of the fundamental frequency. As this is a phenomenon associated with higher acoustic pressures, non-linear propagation of the ultrasound beam occurs mostly in regions where the acoustic pressure is high, i.e. along the beam axis and at the focal position, and not within the low amplitude regions of the beam (e.g. grating lobes and side lobes). These generate sources of noise within the image. By forming the image using only the received second harmonic (non-linear) signal, a less noisy image is generated. In reality, commercial manufacturers utilise a range of different filtering methods to isolate this second harmonic image, which results in a less noisy image.

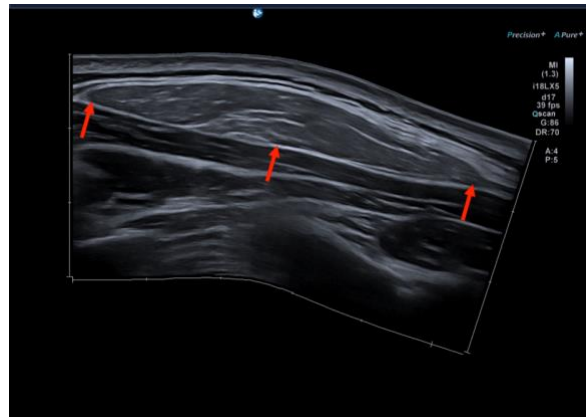
Compounding and Speckle smoothing

In compound scanning, a series of images are acquired at a range of angles and then averaged together to produce an image of mean values. This is of particular importance for displaying curved borders within the body, and for reduction of speckle (noise). Manufacturers typically have their own acronyms for compounding (e.g. "SonoCT", "Aplipure", etc.) and also for image post-processing (e.g. "Precision", "Xres", etc.). The amount of compounding and image post processing can be reduced or increased to suit the preference of the user. Other speckle smoothing techniques, such as frequency compounding, are used by manufacturers in post processing techniques to reduce the overall speckle within an image.

Panoramic Imaging

The ultrasound field of view displayed on a scanner is often limited by the footprint of the transducer. When imaging larger structures, the field of view can effectively be extended by "stitching together" 2D images. Each manufacturer has their acronyms for this, e.g. "Siescape" or just "panoramic". The ability to compose single frames to a panoramic image relies upon the scanner being able to identify structures within the image and is improved by movement of the transducer at a constant speed in one plane [Figure 10].

Figure 10 Panoramic view of a large lipoma (arrows).



DOPPLER ULTRASOUND:

The vascularity of a lesion or the velocity of moving blood within a vessel can be important in making a clinical diagnosis. Ultrasound has the unique ability amongst the other clinical imaging modalities to quantitatively measure blood flow velocities in real time utilising the Doppler technique without the need for injection of contrast.

The Doppler effect

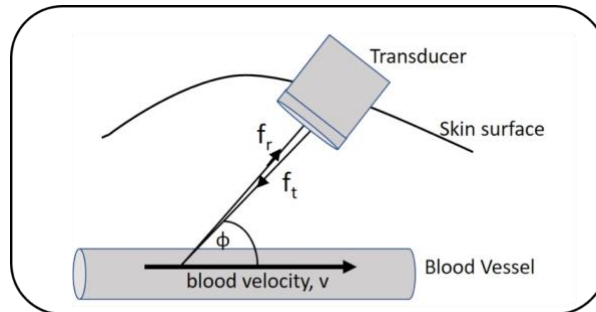
The Doppler shift f_d is equal to the change in frequency of the received ultrasound wave (f_r) relative to the transmitted ultrasound wave (f_t) as a result of the scattering of the transmitted ultrasound wave from red blood cells moving at speed v . Mathematically, f_d is equal to:

$$f_d = f_r - f_t = \frac{2vf_t \cos \phi}{c}$$

where c is the speed of sound of the ultrasound beam in the soft tissue between the transducer and blood vessel and ϕ is the angle between the transmitted ultrasound beam and moving blood [Figure 11].

Figure 11 Blood vessel with blood moving at speed v is insonated by an ultrasound beam of frequency f_t . The angle between the beam and the direction of blood flow is

Φ degrees. Ultrasound is reflected from the moving red blood cells resulting in a Doppler shift so that the frequency of the ultrasound pulse returning to the transducer (f_r) is at a slightly different frequency to f_t .



Consequently, the Doppler shift is not only dependent on the frequency of the transmitted wave but also on the cosine of the angle between the beam and the vessel (angle of insonation).

Rearranging the equation above shows that the speed of blood is equal to:

$$v = \frac{cf_d}{2f_t \cos\phi}$$

Most high-end scanners have spectral, colour and power Doppler options, the controls of which are generally grouped together on one section of the console [Figure 12].

Figure 12 Doppler function keys laid out for ease of the user (arrows).



Pulsed-wave spectral Doppler

In spectral Doppler, the Doppler shift (kHz) or calculated velocity (mm/s or cm/s) is measured within a preselected sample volume and displayed as a continuous function of time scrolling along the bottom of the screen. The magnitude of the Doppler shift (calculated velocity) is displayed as the distance from the baseline (line of zero Doppler shift) such that a positive signal indicates blood moving towards the transducer and a negative signal as blood moving away from the transducer. The brightness of the waveform corresponds to the amplitude of the detected ultrasound at a specific frequency, i.e. number of scatterers moving with a specific velocity.

The gate-length (sample volume or range-gate) and position are selected and modified using a button which usually indicates 'gate' or an icon consisting of 'two parallel bars' and typically varies in size between 1 and 15 mm. The sample volume is typically placed within a vessel using the trackball. The size adjustment depends on whether the user is interested in sampling across the whole vessel or specific locations within the vessel. The size of the sample volume indicates the time during which the scanner will receive and analyse the

signals to determine the Doppler shift. If the sample volume is too large and extends across the vessel boundaries, there will be increased noise in the signal. To ensure that the angle between the ultrasound beam and direction of blood flow is known, the sample volume is aligned either with the vessel walls or with the direction of flow visualised in colour Doppler mode. Calculation of the Doppler frequency from the signal received from the sample volume includes processes of demodulation, high-pass filtering and frequency estimation (FFT processing) which are beyond the scope of this chapter but more information can be found in Hoskins et al 2019. The user has no control over these processes. The user however, does have control over the following features:

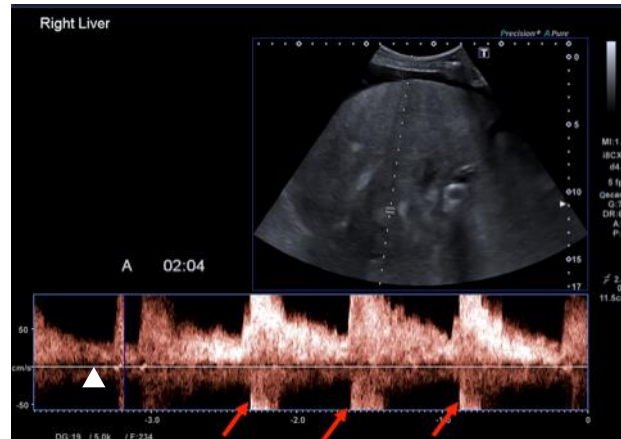
PRF/Scale

The pulse repetition frequency (PRF) (or scale) controls the rate at which pulses are emitted from the transducer with a typical range between 1.1 and 24 kHz. If the PRF is set too low, the Doppler frequency shift is insufficiently sampled (i.e. less than the Nyquist frequency) and “aliasing” will occur such that the high frequencies (velocities) will be incorrectly displayed and visualised as wrapping round on the reverse waveform. If aliasing does occur, the user can increase the PRF, using the ‘control’ button, to correctly display the high velocity signals. If the maximum PRF is reached and aliasing is still present, reducing the transmit frequency or increasing the angle of insonation will further increase the maximum velocity which can be correctly displayed in the spectral trace. Alternatively, high-end scanners may have a ‘high PRF’ mode, which will allow higher velocities to be measured, but knowledge of the depth at which the Doppler signal is being acquired will be compromised and range ambiguity will be introduced. Alternatively, continuous-wave (CW) Doppler can be used to measure very high velocities but the positional depth information on the region from which the high velocities are generated is then lost.

Baseline

The baseline or zero Doppler shift line can be adjusted so that the full Doppler spectrum can be shown especially in instances where there is a large difference in the magnitude of forward and reverse flow. Adjusting the position of the baseline can also prevent aliasing [Figure 13].

Figure 13 Altering the baseline (white arrow head) is one way to reduce “aliasing” (arrows).



High-pass filter

The signal that is processed to provide information on the blood flow velocity will also contain echoes from slow-moving, high-amplitude tissue. The wall filter (high-pass filter) is set so that such echoes are removed from the spectral trace.

Beam-steering angle

If the vessel that is under investigation is parallel to the surface of the skin, even with tilting of the transducer, the angle of insonation can be close to 90° ($\cos \Phi = 0$) resulting in large velocity estimation errors. In such instances, most scanners provide the opportunity to steer the ultrasound beam to either side of the transducer ($\pm 20^\circ$) enabling the required insonation angles of less than 60° to be obtained.

Gain settings

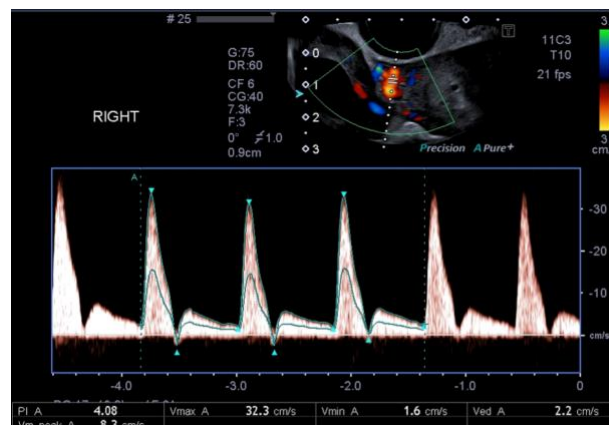
The overall gain of the spectral Doppler display can be adjusted. However, similarly to B-mode imaging, increasing the gain will also increase the background noise of the Doppler

trace. If the gain is increased too much, a duplicate image of the waveform can be reproduced on the reverse image.

Spectral Doppler measurements

The measurements on the spectral Doppler trace are activated via the standard 'measure' button although most premium ultrasound scanners also have an automated measurement function which has its own acronym, e.g. "High Q" or "auto measure", etc. Parameters such as peak velocity, mean velocity, velocity time integral, resistive index and pulsatility index can be calculated from these waveforms [Figure 14].

Figure 14 Automated spectral Doppler Measurement outlining the trace used to obtain the measurements depicted in the box below. Note the sample volume within the colour Doppler box from which the spectral Doppler signal is being obtained.



Colour/Power Doppler

The buttons to select colour and power Doppler modalities are usually placed together and labelled as 'Colour' or 'Colour Doppler Imaging (CDI)' and 'Power' (CPA or PWD) [Figure 12].

Colour Doppler provides information on the direction of blood flow over a relatively large area, whilst power Doppler is more sensitive to regional vascularity. When colour Doppler or power Doppler function buttons are selected, a ROI is overlaid on the grey-scale image - the

position and size of this box can be changed by utilising the trackball and adjacent right and left 'select' keys. The size of the box will have a significant effect on the frame rate as the autocorrelation technique used to calculate the mean velocity in colour Doppler requires that a minimum of two pulses must be transmitted along each line of the image. Premium scanners will utilise more pulses to improve the accuracy of the mean velocity estimation so there is a trade-off between frame rate and the size of the colour Doppler box (i.e. number of lines of data) and the number of pulses used to calculate the mean velocity.

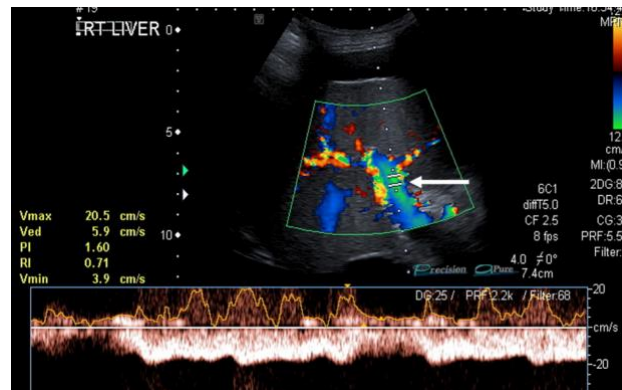
Conventionally in colour Doppler mode, red colour depicts movement of blood towards the transducer and blue indicates movement of blood away from the transducer, such that the colours within the ROI indicate the *mean* velocity at each pixel [Figure 15 and Figure 16]. Colour Doppler is used for assessing the presence of blood flow over relatively large regions of interest. For power Doppler, no directional information on blood flow is obtained but the colour of each pixel indicates the power of the Doppler signal.

Figure 15 Colour Doppler flow showing ROI box, note direction of flow towards probe (red) as indicated by scale on left (arrow).



Figure 16 Portal vein reversal – note spectral gate size (arrow) and flow away from the probe in blue as indicated by the scale. The trace is also below the baseline indicating flow away from the probe. Note the spectral trace is also picking up turbulent flow from the adjacent hepatic artery within the spectral gate. This flow is above the baseline in the spectral Doppler trace indicating hepatopetal

flow and automatic calculations include maximum velocity (Vmax), minimum velocity (Vmin) and Pulsatility index (PI) and Resistive index (RI).



Colour/Power Doppler controls

Many of the controls discussed for optimising the spectral Doppler are also used to optimise colour Doppler imaging. The high-pass filter used to differentiate between tissue and blood is now replaced with a clutter filter to remove the low frequency, high amplitude signals associated with tissue. Mean Doppler frequencies are determined by an autocorrelation technique, rather than the demodulation techniques used in spectral Doppler. The scale of the colour Doppler can be adjusted to ensure that the full range of velocities is displayed in the image. The gain and power output could be optimised to the end that the vessel is full of colour with minimal colour outwith the vessel walls. Owing to the angle dependence of Doppler imaging, the colour or power Doppler ROI box can also be steered using the 'STEER' button to ensure that insonation is not at 90° to the vessel. The Doppler gain can also be adjusted which is usually done with a turn knob on the colour/power Doppler. The optimal setting for scale or pulse repetition frequency (often labelled as 'scale' and 'PRF' respectively) [Figures 13-16] also ensures that aliasing does not occur and that the velocities visualised utilise the entire colour range. These buttons are usually only highlighted for use when Doppler functions are active. In many high-end scanners, the manufacturers have also set a single optimisation key for user ease, which is typically the same as the TGC/Brightness optimisation button.

Microflow/microvascular imaging

Doppler functionality continues to advance and manufacturers are constantly improving the algorithms to increase the sensitivity of Doppler to even slower flow within smaller vessels. Together with a high frame rate and sampling, small vessels with low velocity can now be depicted with minimal motion artefacts and with high resolution resulting in visualisation of vascularity and flow patterns, which have not previously been possible. This feature usually has its own acronym depending on the manufacturer, e.g. “Superb Microvascular Imaging” (SMI), “Microflow imaging” (MFI) or “B- Flow”, etc. While these provide excellent anatomical depiction of the vascularity of the tissue analysed, the quantification of these signals is still not currently available.

Further Reading including additional US techniques

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