

Practical Knobology for Ultrasound-Guided Regional Anesthesia

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Abstract: This article provides an instructive review of the essential functions universal to modern ultrasound machines in use for regional anesthesia practice. An understanding of machine knobology is integral to performing safe and successful ultrasound-guided regional anesthesia.

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A thorough understanding of machine “knobology” is essential to the safe and successful practice of ultrasound (US)-guided regional anesthesia. Much like the basic functions of our television sets (eg, volume, brightness, contrast, sharpness), all US machines share the same operative features that can and should be manipulated to optimize imaging. These basic features are frequency, gain, depth, focus, time gain compensation (TGC), color Doppler, power Doppler, compound imaging, and, on some models, tissue harmonic imaging. A brief description of each is provided below. Additional information and complementary figures relating to this article can be found in 2 previously published review articles,^{1,2} which are reprinted as resource material in the present special supplement issue of *Regional Anesthesia and Pain Medicine*.^{1,2}

Frequency

Selecting the appropriate frequency of the emitted US wave is perhaps the most important of all adjustments made by the operator. The frequency range most commonly used for peripheral nerve blockade is between 3 and 15 MHz. Higher frequencies provide superior axial resolution but this is at the expense of increased beam attenuation and reduced penetration (Fig. 1). Axial resolution refers to the ability of the US machine to distinguish between 2 objects at different depths in line with the axis of the beam. High-frequency transducers are usually best for target depths of up to 3 to 4 cm. For deeper targets, a lower-frequency transducer is often necessary. The operator must therefore select the transducer that strikes the optimal balance between the highest possible frequency and tissue penetration to the appropriate depth. On most modern US machines, frequency selection may be considered a 2-step process consisting of a coarse and then fine adjustment. The first “coarse”

frequency adjustment relies on choosing the appropriate transducer according to the expected depth of the target. Transducer categories can be divided into high- (8–12 MHz), medium- (6–10 MHz), and low- (2–5 MHz) frequency ranges. Once the appropriate transducer is selected, the operator can then fine tune the frequency of the US wave emitted from the transducer by actively selecting only the upper, mid, or lower frequencies within each transducer’s frequency range. This fine adjustment is made by selecting an actual megahertz value on some US machines (eg, Philips Healthcare, Andover, MA) or by toggling the RES (resolution), GEN (general), or PEN (penetration) button on other machines (eg, Sonosite, Bothell, WA), which correspond to the upper, mid, or lower frequencies of the transducer’s range.

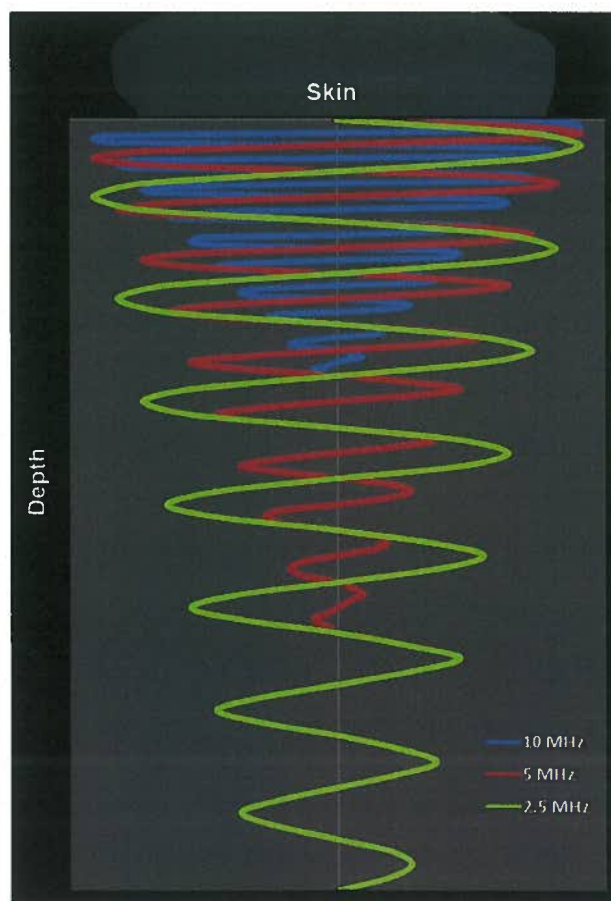


FIGURE 1. Attenuation (energy loss) is directly proportional to the frequency of the sound waves and the distance that the sound waves must travel. Note how the lower-frequency US waves are less attenuated compared with the higher-frequency (10 MHz) wave at any given distance (depth).

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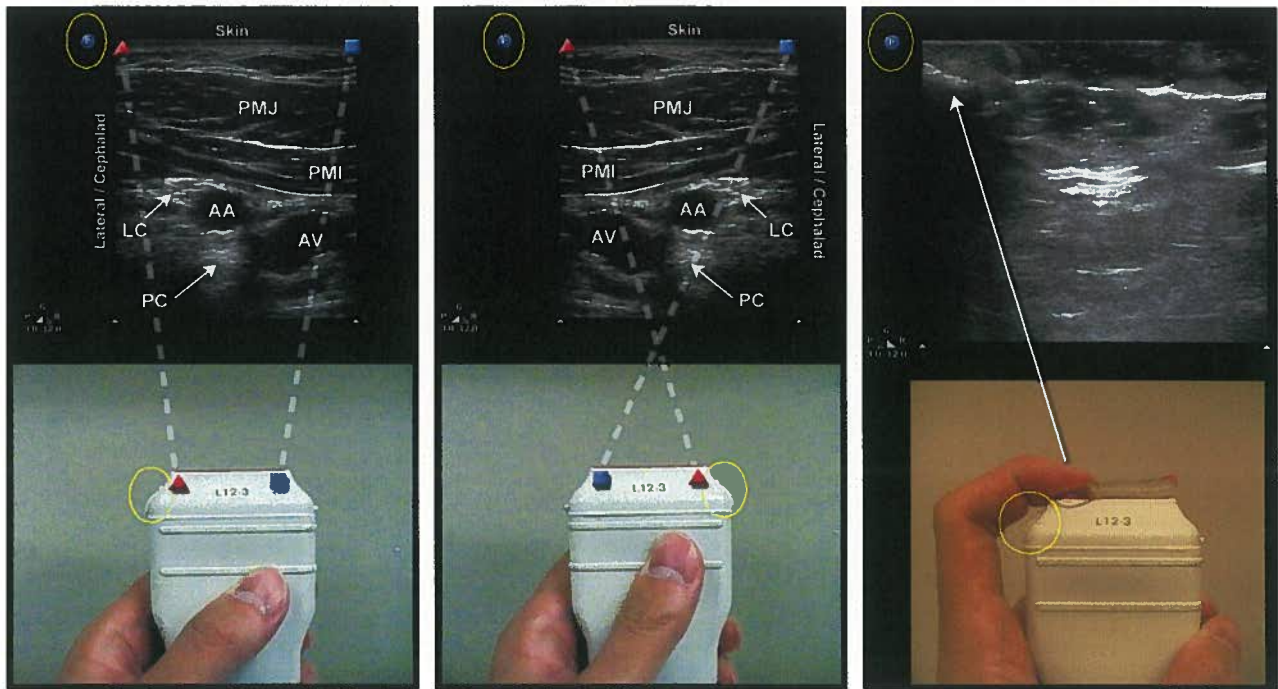


FIGURE 2. Transducer orientation. Transducer selection is complete with an understanding of transducer orientation relative to the corresponding image on the US screen. Regardless of manufacturer, every transducer has a fixed label (eg, palpable groove, notch, or prominence) on one end that corresponds to an adjustable label (eg, blue dot or manufacturer’s insignia) on the left or right side of the US screen. Left and middle, Short-axis image of the infraclavicular brachial plexus. Position of labels on transducer and screen are same. Middle, Position of labels on transducer and screen are opposite. AA indicates axillary artery; AV, axillary vein; LC, lateral cord; PC, posterior cord; PMJ, pectoralis major muscle; PMI, pectoralis minor muscle. Right, To determine which end of the transducer corresponds to the left or right side of the 2-dimensional image projected on the screen, the provider is advised to tap one end of the transducer before commencement of scanning.

Gain

The gain dial allows the operator to change how bright (hyperechoic) or dark (hypoechoic) the image appears. The mechanical energy of the returning sound waves (echoes) is converted by the US machine into an electrical signal, which in turn is converted into a displayed image. Increasing the gain amplifies the electrical signal produced by these echoes, which causes an increase in the brightness of the entire image, including background noise, and the potential for artifact generation (Fig. 3, middle), whereas too little gain can negate real echo information (Fig. 3, right). Increasing the gain also reduces lateral resolution (see below).

Time Gain Compensation

Similar to the gain dial, the TGC control panel allows the operator to make adjustments to the brightness, but unlike the

gain dial that increases the brightness for all points in the field, TGC allows the operator to adjust the brightness independently at specific depths in the field (Fig. 4). Time gain compensation may be considered similar to an equalizer on a stereo. To fully appreciate TGC, one must recall the principle of attenuation. As US waves pass through tissues, they lose energy (amplitude) owing to scatter, reflection, and absorption. Attenuation varies depending on both the beam frequency and the tissue through which the US waves travel. The latter is represented by the attenuation coefficient, which differs between tissue types. Attenuation also varies directly with depth of penetration. If the US machine were to display the actual amplitude of echoes returning to the probe, the image would be progressively hypoechoic from superficial to deep. Although US machines are designed to automatically compensate for attenuation, it is not always accurate, allowing for manual adjustment of TGC. The



FIGURE 3. Left, Correct gain setting. The target nerves of the infraclavicular brachial plexus (LC indicates lateral cord; PC, posterior cord; medial cord not in view), axillary artery (AA), axillary vein (AV), and surrounding musculature are clearly visible. Middle, The overall gain is set too high. Right, The overall gain is set too low.

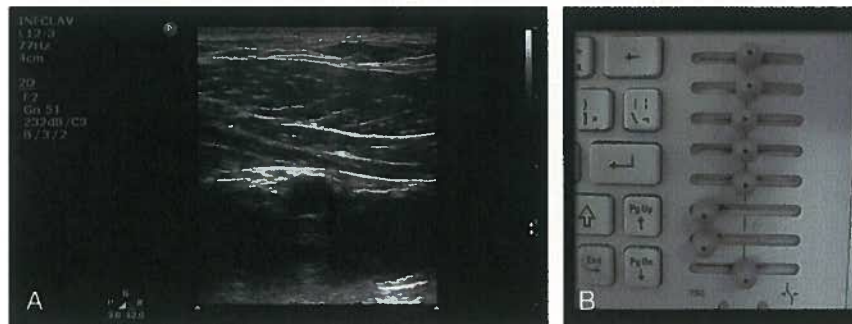


FIGURE 4. Incorrect TGC setting. A, The hypoechoic band in the middle of the image makes it difficult to identify the posterior cord. This is because the TGC dial corresponding to this depth of the field is turned down (B) creating a band of undergain.

TGC panel is most commonly used to increase the brightness of structures in the far field (deep structures) to create a uniform image. Some machines have individual controls (“slide pots”) for each small segment of the display (eg, Philips, GE), whereas others simply have “near” and “far” gain (eg, Sonosite).

Depth

The depth of the image must be adjusted so that the target structure falls within the field of view (Fig. 5, top). The objective of depth adjustment is to set the depth to just below the structure of interest. Selecting a greater depth than necessary results in a smaller target owing to the change in aspect ratio of the displayed image (Fig. 5, bottom left), whereas selecting too shallow a depth can obscure the target and other important structures (Fig. 5, bottom right). Setting the appropriate depth

also optimizes temporal resolution. Temporal resolution can be thought of as the frame rate and refers to the rate at which images are produced (expressed in frames per second). Temporal resolution depends on the rate at which successive US waves are emitted to form a full sector beam. Conceptually, US waves are emitted in pulses, the next one emitted only when the previous one has returned to the transducer. For deeper structures, this rate is slower by definition. Temporal resolution is thus forfeited as depth is increased. However, US machines can preserve temporal resolution by reducing the width of the sector beam as the depth is increased. Reducing the sector width effectively reduces the number of emitted waves that must return to the transducer, thereby reducing the time before an image is displayed. Temporal resolution is of more importance when visualizing moving objects such as during cardiac imaging

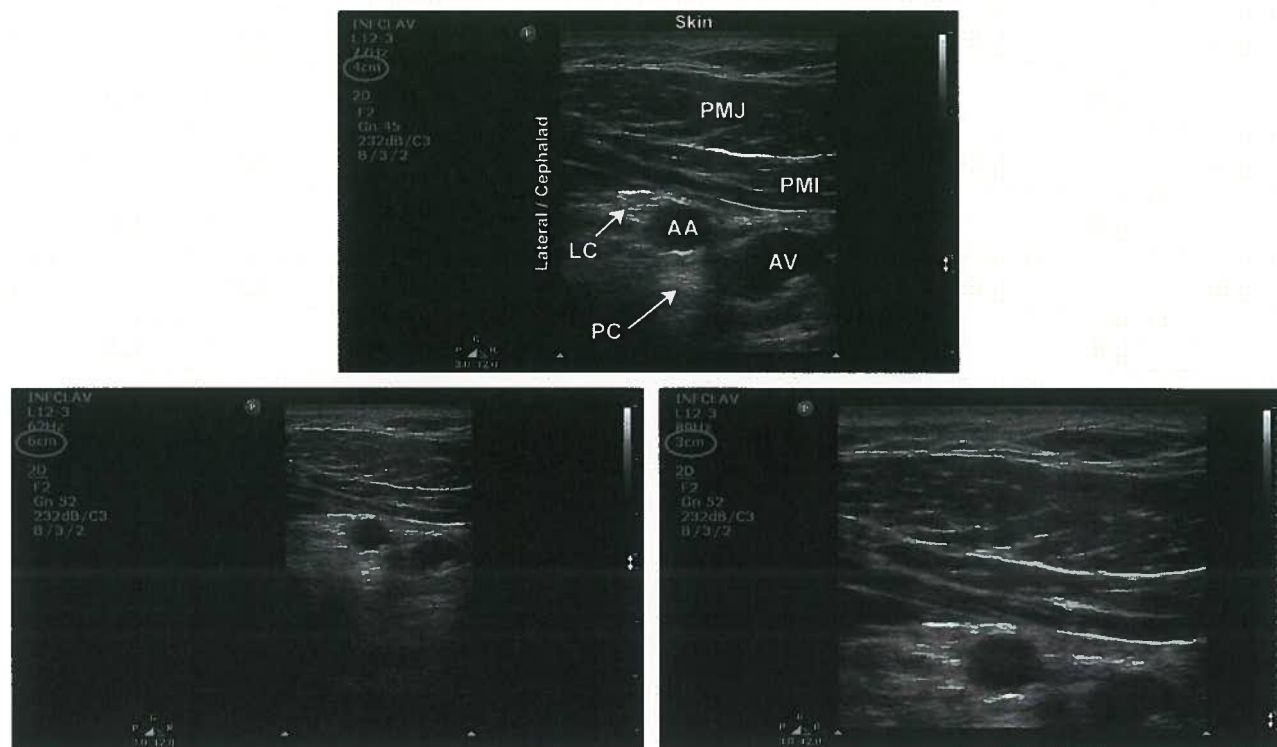


FIGURE 5. Top, Correct depth setting. The lateral cord (LC), posterior cord (PC), axillary artery (AA), axillary vein (AV), and surrounding musculature are clearly visible in their entirety. Bottom Left, Excessive depth setting. The depth has been set inappropriately deep, decreasing the size of the target structures displayed on the US screen. Bottom Right, Inadequate depth setting. The depth needs to be increased to view the region posterior to the axillary artery where one would expect to find the posterior cord.

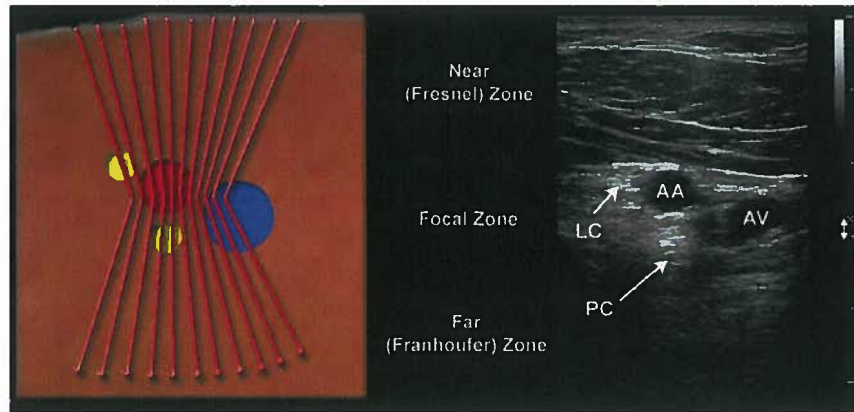


FIGURE 6. The focal zone is the point where the beam is narrowest. Lateral resolution is optimal in this zone, before beam divergence. Lateral resolution refers to the ability of the US machine to correctly display 2 objects lying at the same depth as separate structures.

and arguably less important in regional anesthesia. A low frame rate, however, could create a blurred image during needle movement or rapid injection of local anesthetic.

Focus

Adjusting the focus can improve lateral resolution. Lateral resolution refers to the machine’s ability to distinguish 2 objects lying beside one another at the same depth, perpendicular to the US beam. Multiple piezoelectric elements arranged in parallel on the face of the transducer emit individual waves that together produce a 3-dimensional US beam. The 3-dimensional US beam first converges (Fresnel zone) to a point where the beam is narrowest (focal zone) and then diverges (Fraunhofer zone) as

it propagates through the body (Fig. 6). As the beam diverges, the individual element waves no longer travel in parallel and become further spaced apart from one another. Ideally, each individual element wave would strike (and, consequently, produce a corresponding image) every tissue in the field, no matter how close 2 tissues lie next to one another. Conceptually, target tissues may be missed by “slipping in between” the incoming element waves if the parent beam is divergent. The focus dial allows the operator to cinch the parent beam at various depths in the field to limit the amount of beam divergence, thereby improving lateral resolution at the level of the target. The focus level is generally represented by a small arrow at the left or right of the image. It is important therefore to position the focus

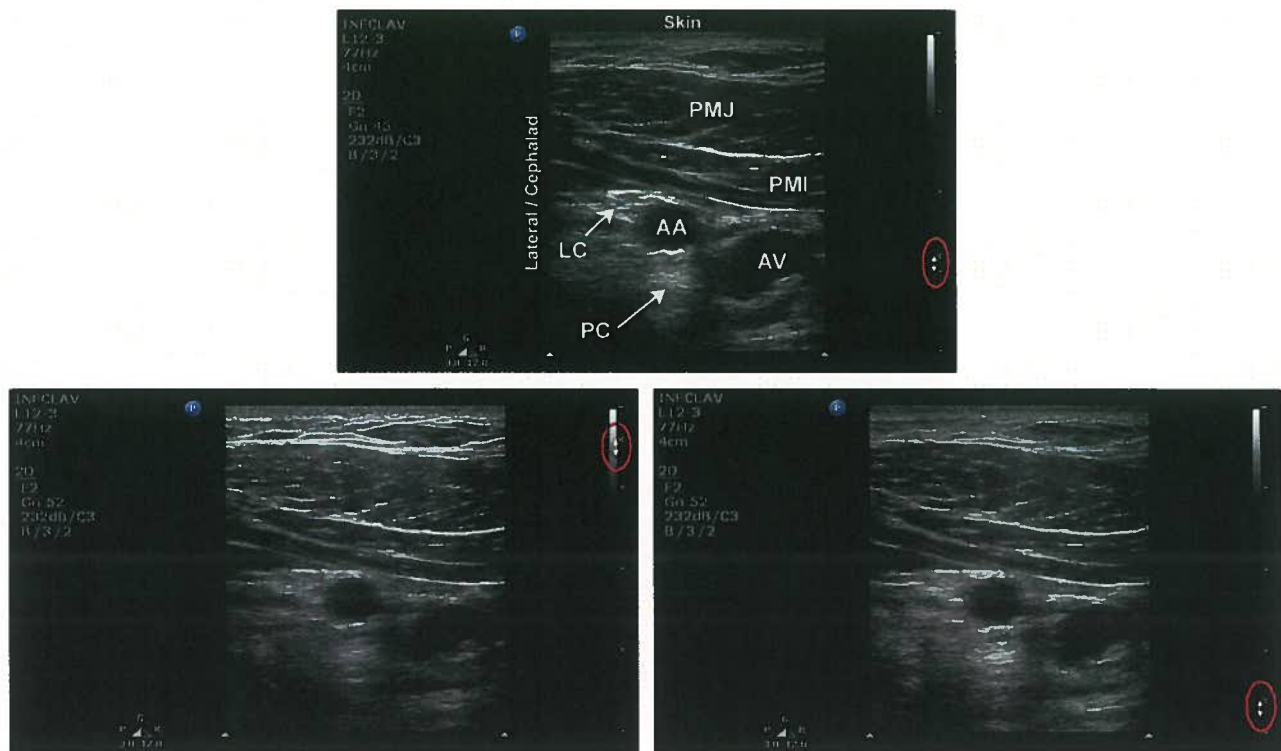


FIGURE 7. Top, Appropriate focus setting for imaging the infraclavicular brachial plexus. Bidirectional arrows along the right border of the image designate the level at which the focus is set. Bottom Left, Lateral resolution at the level of the nerves is now degraded because of inappropriately shallow focus setting. Bottom Right, Inappropriately deep focus setting.

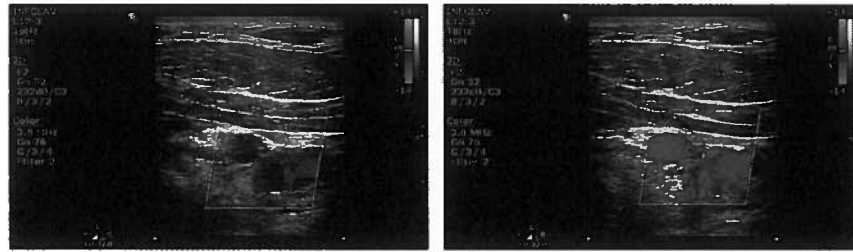


FIGURE 8. The effect of tilting a probe on the short-axis view of the axillary artery and vein. Left, There appears to be no flow as the angle of the beam is at 90 degrees to the direction of blood flow. Right, Slight tilting of the probe to change the angle of insonation results in visualization of blood flow.

of the beam at the same level as the target(s) of interest (Fig. 7). Some machines offer the ability to set multiple focus zones. However, increasing the number of focal zones occurs at the expense of temporal resolution as the machine spends more time listening for returning echoes and processing each image.

Color Doppler

Color Doppler technology superimposes Doppler information on the real-time image and allows for the identification and quantification (velocity, direction) of blood flow. The Doppler principle states that if a sound wave is emitted (or reflected) by a moving object, the frequency of that sound wave will change as the object moves toward or away from a stationary listener. This is the same principle that explains why the pitch (frequency) from a siren on an approaching ambulance sounds high and then low once the ambulance passes and speeds away.

If a US wave is emitted from a stationary transducer and strikes moving red blood cells, the reflected US wave will have a different frequency from the original emitted wave. Blood moving away from the transducer will return at a lower frequency than the original emitted wave and is represented by blue. Blood moving toward the transducer will return at a higher frequency than the original emitted wave and is represented by red. It is important for inexperienced users to recognize that red does not necessarily correlate with arterial or blue with venous blood. The change in frequency is known as the Doppler shift, and it is this principle that can be used in cardiac and vascular applications to calculate both blood flow velocity and blood flow direction. The Doppler equation states that:

$$\text{Frequency shift} = (2VF_t)(\cos \alpha)/c$$

where V is the velocity of the moving object, F_t is the transmitted frequency, α is the angle of incidence between the US beam

and the direction of blood flow, and c is the speed of US in the blood. It is also important to note that the cosine of 0 degrees is 1 and the cosine of 90 degrees is 0. Therefore, as the angle of incidence approaches 90 degrees, large errors are introduced into the Doppler equation (Fig. 8). However, the most important application of Doppler technology for the regional anesthesiologist is to *confirm the absence* of blood flow in anticipated trajectory of the needle, rather than the quantification of the actual velocity or direction of this flow. On some US machines, the amount of color Doppler displayed (sensitivity) is adjusted by turning the gain knob, whereas other machines offer a separate knob. However, increasing the sensitivity can result in the production of motion artifacts (false-positive). Changing the Doppler angle, altering the frequency (based on the previous equation), or altering the Doppler scale can also affect Doppler sensitivity. Color Doppler requires more processing time per line of field information compared with simple B-mode imaging, so image quality is visibly reduced, although this varies greatly between different US manufacturers.

Power Doppler

Power Doppler is a newer US technique that is up to 5 times more sensitive in detecting blood flow than color Doppler. Power Doppler can detect vessels that are difficult or impossible to see using standard color Doppler. A further advantage is that, unlike color Doppler, power Doppler is almost angle-independent. Such benefits, however, come at the expense of more motion artifact with subtle movements such as respiration. Furthermore, power Doppler cannot detect the direction of flow.

Compound Imaging

Compound imaging electronically steers the transducer array to rapidly capture several (3–9) overlapping scans of the same

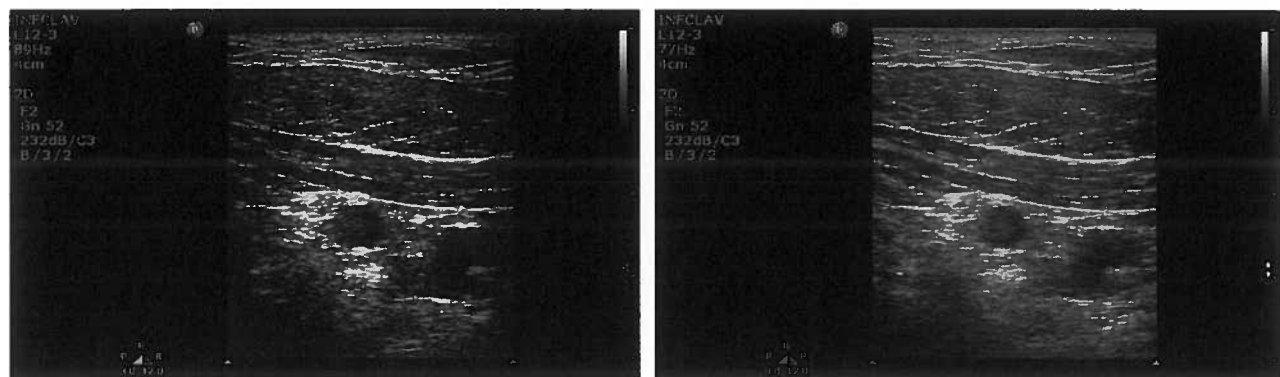


FIGURE 9. Left, Compound imaging turned OFF. Right, Compound imaging turned ON. Note the increased speckle artifact and decreased resolution in the left panel compared with that in the right panel.

tissue from different view angles. The sonographic information from these different angles of insonation is electronically combined to produce a single image (spatial compound imaging). Compound imaging improves image quality compared with conventional US by reducing speckle and other acoustic artifacts and by improving the definition of tissue planes (Fig. 9). Frequency compound imaging uses differing frequencies to create a single image.

Compound imaging technology is designated by different names depending on the US manufacturer. For example, it is called SonoMB on Sonosite machines, CrossXbeam on the GE Logic range, and SonoCT with Philips equipment.

Tissue Harmonic Imaging

Echoes returning from the target to the US transducer include those from the target as well as a variety of echoes from other scattered points in the field, which decreases image quality. When echoes are reflected, they are reflected not only at the original frequency but also at various multiples of the original frequency. Tissue harmonic imaging, if available, preferentially includes the higher- (harmonic) frequency echoes because these produce less scatter and artifact and therefore a higher resolution

image is obtained. Although tissue harmonic imaging seems to particularly improve visualization of hypoechoic, cystic structures, it can worsen needle visibility.

Optimization Button

Many newer machines now incorporate an automatic all-in-1 image optimization button that allows the US machine to instantaneously combine many of the aforementioned features to create the ideal image. This can be a simple and effective way to improve the quality of the image, and it does not limit the provider from making further manual adjustments thereafter.

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