Breast ultrasound knobology and the knobology of twinkling for marker detection

Christine U. Lee[1]^, Gina K. Hesley[1]^, Taylor A. Pierson[1], Rebecca L. Higgins[1], Matthew W. Urban[2,3]^

[1]Department of Radiology, Division of Breast Imaging and Intervention, Mayo Clinic, Rochester, MN, USA; [2]Department of Radiology, Division of Radiology Research, Mayo Clinic, Rochester, MN, USA; [3]Department of Physiology and Biomedical Engineering, Mayo Clinic, Rochester, MN, USA

Contributions: (I) Conception and design: CU Lee, GK Hesley, MW Urban; (II) Administrative support: None; (III) Provision of study materials or patients: All authors; (IV) Collection and assembly of data: All authors; (V) Data analysis and interpretation: CU Lee, GK Hesley, MW Urban; (VI) Manuscript writing: All authors; (VII) Final approval of manuscript: All authors.

Correspondence to: Christine U. Lee, MD, PhD. Department of Radiology, Division of Breast Imaging and Intervention, Mayo Clinic, 200 First St. SW, Rochester, MN 55905, USA. Email: lee.christine@mayo.edu.

Abstract: Breast ultrasound utilizes various scanning techniques to acquire optimal images for diagnostic evaluation. During interventional procedures, such as ultrasound-guided biopsies or preoperative localizations, knowledgeable and purposeful scanning adjustments are critical for successfully identifying the targeted mass or biopsy marker or clip. While most ultrasound scanning parameters are similar across different vendors, detailed descriptions specifically addressing the scanning parameters—often referred to as "knobology"— for breast ultrasound is at best limited in the literature. This review highlights ten key operator-controlled adjustments (including transducer selection, beam focusing, power, depth, gain and time gain compensation, harmonic imaging, spatial compounding, dynamic range, beam steering, and color Doppler) that significantly influence image quality in breast ultrasound. Each adjustment is accompanied by an "In practice" section providing examples and practical tips on implementation. The last topic discusses color Doppler which is generally used in breast ultrasound for evaluating the vascularity of a finding. Color Doppler, or more specifically, color Doppler twinkling, can be leveraged as a technique to detect certain breast biopsy markers that are challenging to detect by conventional B-mode ultrasound. While the cause of color Doppler twinkling is still under active investigation, twinkling is a clinically well-known, compelling ultrasound feature typically described with kidney stones. A step-by-step guide on how to use color Doppler twinkling to detect these markers is provided.

Keywords: Breast ultrasound; knobology; color Doppler twinkling; twinkling; ultrasound scanning

Received: 19 June 2024; Accepted: 14 October 2024; Published online: 29 October 2024.
doi: 10.21037/tbcr-24-30
View this article at: https://dx.doi.org/10.21037/tbcr-24-30

Introduction

Background

Fourteen years before X-ray devices using molybdenum tubes instead of tungsten tubes ushered in the era of modern mammography units (1), Dr. J. J. Wild from the Department of Surgery at the University of Minnesota first described using high-frequency ultrasound (then called an ultrasonoscope) to image the breast in 1951 (2). Technological advancements in ultrasound hardware and software over the ensuing decades helped to establish its diagnostic value. As early as forty years ago, ultrasound

^ ORCID: Christine U. Lee, 0000-0001-5298-6949; Gina K. Hesley, 0000-0003-1246-696X; Matthew W. Urban, 0000-0003-1360-4287.

Table 1 Topics covered

Section	Topic
I	Transducers
II	Beam focusing
III	Power
IV	Depth
V	Gain and time gain compensation
VI	Harmonic imaging
VII	Spatial compounding
VIII	Dynamic range
IX	Beam steering
X	Color Doppler

was recognized for its effectiveness in guiding procedures and assessing breast lesions preoperatively (3). Today, breast ultrasound is a standard-of-care diagnostic imaging modality in breast imaging. While ultrasound technology continues to improve, the quality of the ultrasound images remains largely operator-dependent.

Rationale and knowledge gap

In 2003, the risk assessment tool BI-RADS® (Breast Imaging Reporting And Data System), developed by the American College of Radiology (ACR), incorporated breast ultrasound to standardize terminology and communication of findings (4). Accurate description of ultrasound findings using the BI-RADS lexicon depends on the quality of the images acquired. While some parts of image quality may be vendor-dependent, many scanning parameters are vendor-agnostic and can be generally described, leaving it up to the operator to improve image quality based on the underlying science.

Objective

The purpose of this manuscript is to present a high-level approach to obtaining high-quality breast ultrasound images by describing the vendor-agnostic components and available adjustable scanner parameters (here, loosely called "knobs" encompassing both physical and touch-screen versions) when performing breast ultrasound. Additionally, color Doppler ultrasound, mentioned only once in the 2021 ACR Practice Parameter for the Performance of a Breast Ultrasound Examination (5), in the context of characterizing the internal vascularity of the lesion, has been used to evaluate the color Doppler twinkling properties of some breast biopsy markers (6-8). This manuscript includes a description of the knobology of color Doppler twinkling to aid in ultrasound detection of some breast biopsy markers.

Knobology

Ten knobology topics are discussed (*Table 1*), and for each topic, an "In practice" section provides examples and practical tips on implementation.

(I) Transducers

The ultrasound transducer transmits sound waves generated by a voltage applied to piezoelectric crystals and detects returning sound waves. Each transducer has a preferred resonant frequency and bandwidth (range of frequencies). The 2021 ACR Practice Parameter for the Performance of a Breast Ultrasound Examination recommends using a high-resolution, real-time, linear-array, broad-bandwidth transducer operating at a center (resonant) frequency of at least 12 MHz, preferably higher (5). The recommendation for a higher center frequency is application-driven because the breasts are relatively superficial structures compared, for example, to the liver and gallbladder where a lower center frequency is more appropriate. The recommendation for a broad bandwidth describing a broader range of transmitted and received frequencies provides the benefits of higher resolution with higher frequencies and deeper tissue penetration with lower frequencies, all with the same transducer.

Transducers are primarily characterized by the range of sound wave frequencies (MHz), emitted from the transducer. The frequencies of the sound waves are determined by the properties of the piezoelectric crystals that make up part of the transducer. A high-frequency transducer typically used for breast ultrasound may operate at a frequency range of 5–15 MHz. On the scanner console, there is often a knob that allows the user to select a center frequency within the range of frequencies allowed by the transducer. The sound waves leave the transducer in short bursts or pulses at a rate characterized by the pulse repetition frequency (PRF) or number of pulses emitted from the transducer in one second (Hz).

One of the first steps in acquiring high-quality images is to understand which transducer may be best suited for the task at hand. Ultrasound image resolution of the desired imaging target is determined primarily by the transducer's

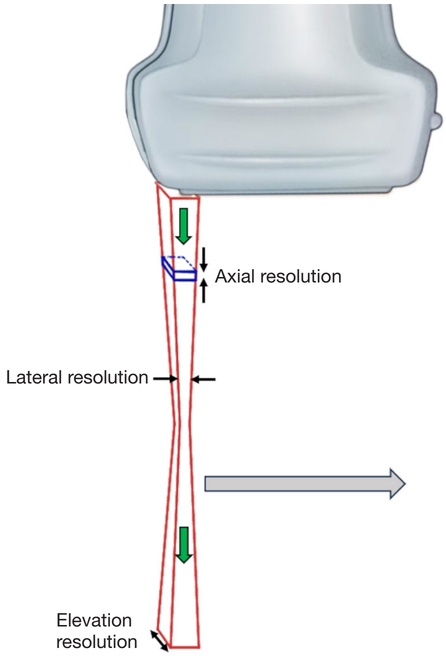


Figure 1 Axial and lateral resolutions. The axial resolution, which is determined by the transmit frequency associated with the array of the piezoelectric crystals, is generally better than the lateral resolution, which is determined by the footprint, application, and transmit frequency. The ultrasound beam (red) emitted along its path (green arrows) from the transducer gets narrowed at the focal zone and then widens with more depth as it sweeps from one end of the transducer to the other (large gray arrow). The axial resolution (blue), equal to half the spatial pulse length, defines to what extent one can separate two close objects or points along the direction of the beam. The lateral resolution varies slightly along the beam's path. The elevation resolution defines the slice thickness of the imaging plane.

frequency and footprint (linear, curvilinear, or phased array). The axial and lateral resolutions (LRs) of the image acquisitions (*Figure 1*) primarily determine the in-plane resolution of the image, while the out-of-plane resolution or "slice thickness" is determined by the elevation resolution. The axial resolution is determined by the spatial pulse length (SPL), which is a function of frequency, f, and pulse length in terms of N cycles (SPL = $N\lambda$ = Nc/f where c is the speed of sound in tissue, commonly assumed as 1,540 m/s and λ is the ultrasound wavelength). The LR is also related to the wavelength and the size of the aperture or footprint used to transmit and receive. Linear transducers are favored for more superficial imaging, such as the breasts, while curvilinear transducers, which have a larger footprint or aperture compared to linear transducers, are often used for deeper structures, such as those in the abdomen or pelvis. The increased lateral coverage afforded with a curvilinear transducer compared to a linear transducer is different from its LR which is still dependent on wavelength or frequency (9). The relationship is fundamentally described by the following equation:

$$LR = 1.4\lambda \frac{Focal\ length}{Aperture} = 1.4\lambda \frac{z_f}{D} \quad [1]$$

where z_f is the focal depth and D is the aperture width.

On some ultrasound scanners, there is an option to turn on virtual convex imaging for linear transducers; this expands the field of view by post-processing acquired signals or employing vendor-specific steering.

In practice

Transducers with lower transmit frequencies send and receive signals that can penetrate through more tissue at the cost of lower axial resolution (*Figure 1*), but this may be useful in certain applications, for instance, when imaging breasts with more than 3 cm of tissue while in the supine position, deep lesions, lactating breasts, mastitis, inflammatory breast cancer, or dense breasts. High transmit frequencies (e.g., >12 MHz) mean higher axial resolution at the cost of less tissue penetration, but this may be ideal, for example, for imaging superficial lesions, interrogating the nipple-areolar complex, or locating superficial findings such as some biopsy markers.

When evaluating a possible sebaceous cyst or epidermal inclusion cyst and looking for the tract in the skin, using an even higher frequency transducer, 18 or 24 MHz, for example, may be helpful for identifying the tract in the skin or for better characterizing the finding (*Figure 2*). High-frequency transducers can sometimes be used to assess for the presence of a sonographically visible marker (clip) in a resected specimen, such as a positive lymph node, in the operating room while waiting for the specimen radiograph.

(II) Beam focusing knobs

The ultrasound beam, which is affected by many factors, refers to the collection of acoustic waves emanating from the piezoelectric crystals. Using beam-focusing knobs, the operator can narrow the ultrasound beam and improve the resolution in that area. Because the axial resolution is inherently better than the LR in ultrasound, narrowing

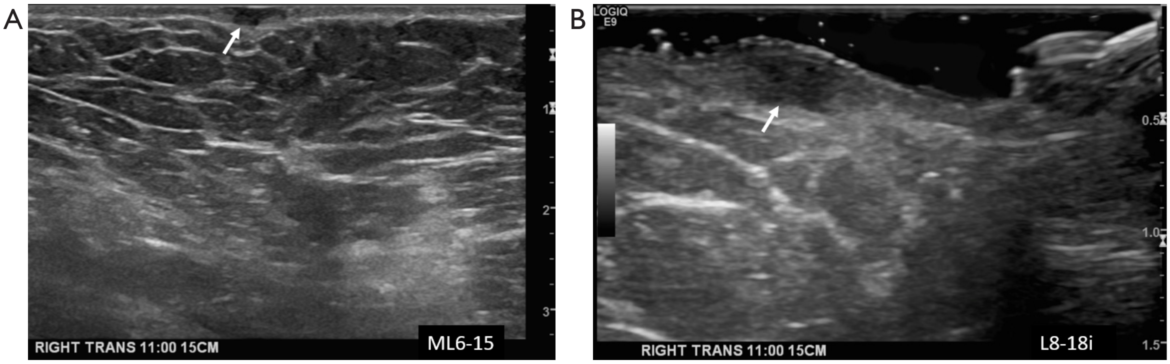


Figure 2 Selection of transducer. A conventional high-frequency transducer for breast ultrasound (ML6-15, GE Logiq E9, GE Healthcare, Wauwatosa, WI, USA) shows a skin finding (A, arrow). Using a higher frequency transducer, such as the L8-18i (B), the details of the skin finding (B, arrow), such as the margins, can be improved. Some of these features are vendor-specific. In this case, the higher frequency transducer provides higher axial and lateral resolution, but there is an overall smoothed pixelated appearance, making it somewhat blurrier than the lower frequency transducer. This could be a vendor-specific observation and can be mitigated by using harmonic imaging.

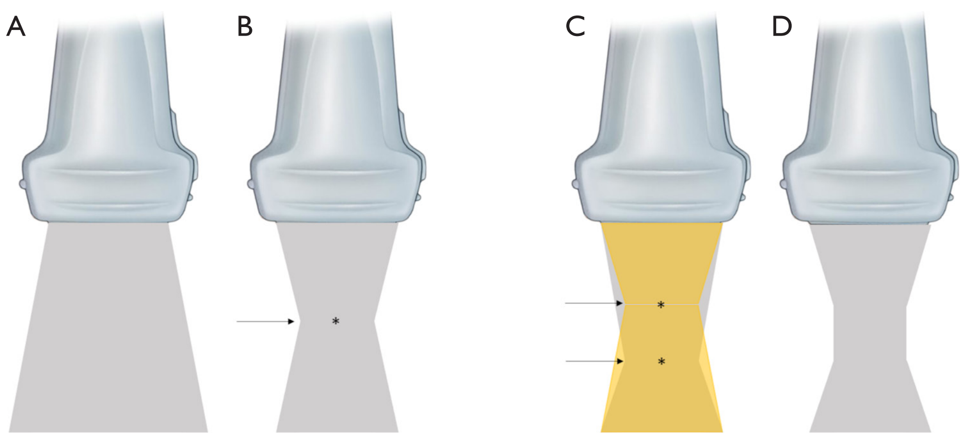


Figure 3 Focal zones. To improve the lateral resolution of the image, the ultrasound beam (A, gray) can be focused (B, arrow) to improve the lateral resolution at that location (*). Multiple beam-focusing placements (C, arrows) cause an effective column of improved lateral resolution (D), which may improve feature sharpness. (Used with permission by Mayo Foundation for Medical Education and Research, all rights reserved. Original image modified by Christine Lee).

the ultrasound beam improves the LR more than the axial resolution of the image. Beam-focusing knobs allow the operator to choose a single focus of beam narrowing or create a column or effective focus of narrowing by using multiple foci (*Figure 3*) and combining the results from multiple pulse-echo acquisitions. For some vendors, the distance between the foci of narrowing can be specified by the user. Regardless of the combination of adjustments made, the overall effect is to improve the resolution of the image at the area(s) of beam narrowing, generally referred to as the focal zone; the LR is further improved with higher transmit frequencies. One of the trade-offs of using multiple focal points is a decrease in frame rate, which is generally minimal and may not be readily noticeable.

In practice

Adjusting the focal zones is generally an automatic step in breast ultrasound, particularly when imaging a particular finding. Whether one focal zone or multiple focal zones are used depends on the case at hand. Spending time to cycle through a few focal zones can sometimes provide more information on the finding (*Figures 4,5*). For very superficial findings, such as those in the skin, a thick layer of gel may help bring the finding to the level of the most superficial

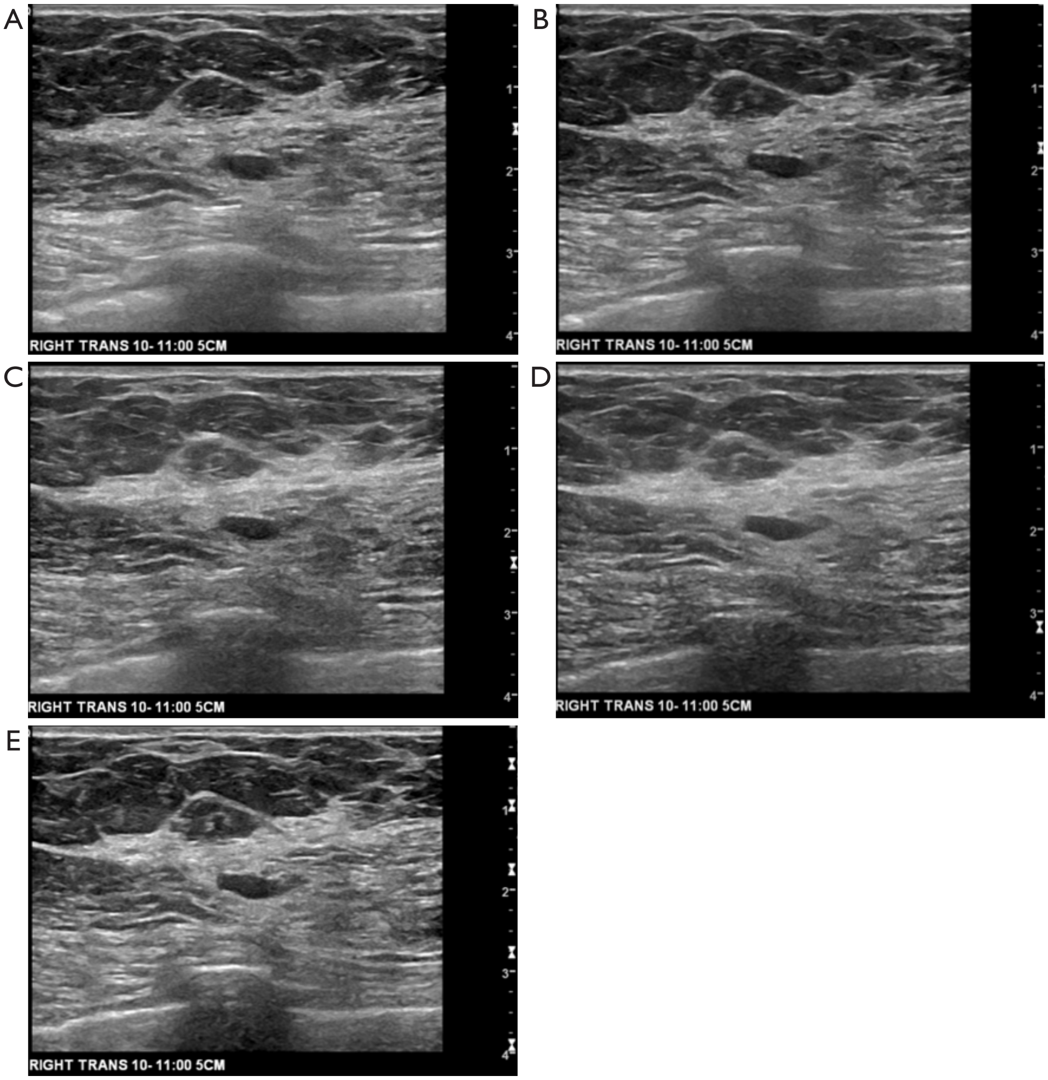


Figure 4 Placement of single focal zones and multiple focal zones. Changing the placement of a single focal zone [A-D hourglass (⧖) symbol on the right-hand side of each image] while keeping all other imaging parameters the same shows subtly improved sharpness near the level of each focal zone. For example, when the focal zone is placed at 1.50 cm deep (A), the tissues at this depth are sharper than when the focal zone is at 3.25 cm (D). In (A-D), each image with a single focal zone had a frame rate of 64 fps, but in (E) with multiple focal zones, the frame rate dropped to 16 fps.

focal zones (*Figure 6*). This technique may also be helpful when scanning the nipple-areolar complex or near surgical scars with keloid formations.

In practice, knob adjustments may not result in appreciable diagnostic changes (*Figure 7*), likely due to the variable sonographic features of suspicious breast findings. Until the pathologic changes contributing to sonographic characteristics are better understood, it is advisable to briefly adjust the knobs such as focal zones, as this may provide additional diagnostic information. One tip to consider when evaluating deeper structures is compressing the transducer a little more, effectively dispersing overlying tissues and suspensory ligaments that can shadow the deeper tissues. This tip, combined with optimizing the focal zone, can sometimes produce sharper images of a deep finding.

(III) Power knob

Ultrasound waves emitted from the transducer are a source of acoustic energy measured in Joules (J), and the rate at

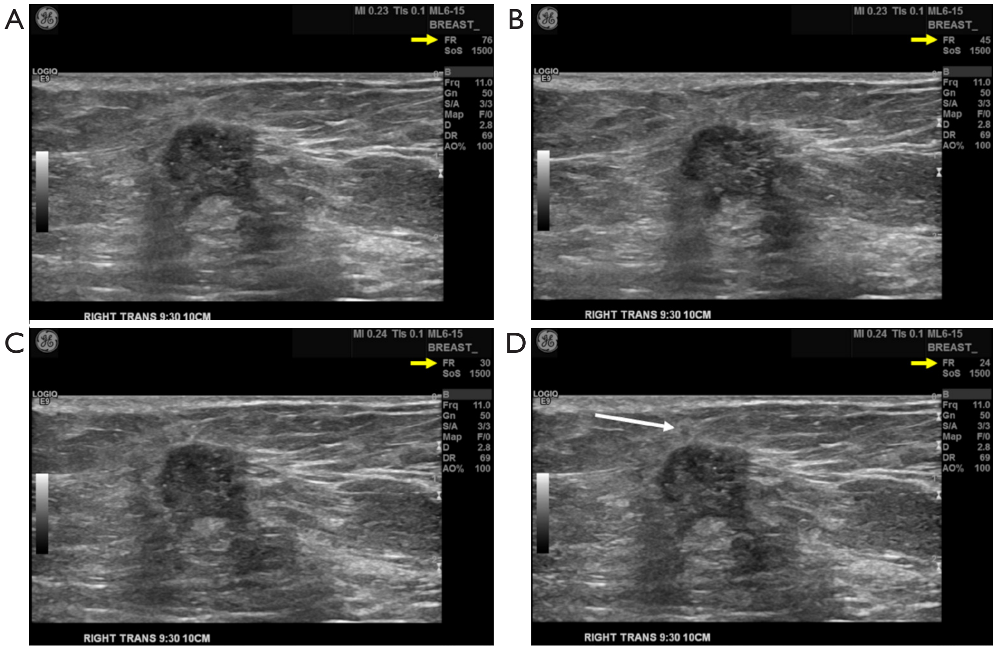


Figure 5 Multiple focal zones. Using an ML6-15 transducer and keeping all other parameters the same, images of this irregular hypoechoic mass were acquired with one to four focal zones [A-D, hourglass (⧖) symbols on the right-hand side of each image]. Imaging differences are subtle with (B) slightly off plane compared to (A,C,D). Utilizing multiple focal zones can help better evaluate tissue characteristics, e.g., a subtle, small, taller-than-wide, isoechoic-hypoechoic mass with irregular and indistinct margins (D, white arrow) when a focal zone near the near field is placed. Note that increasing the focal zones decreases the frame rate (A-D, yellow arrows).

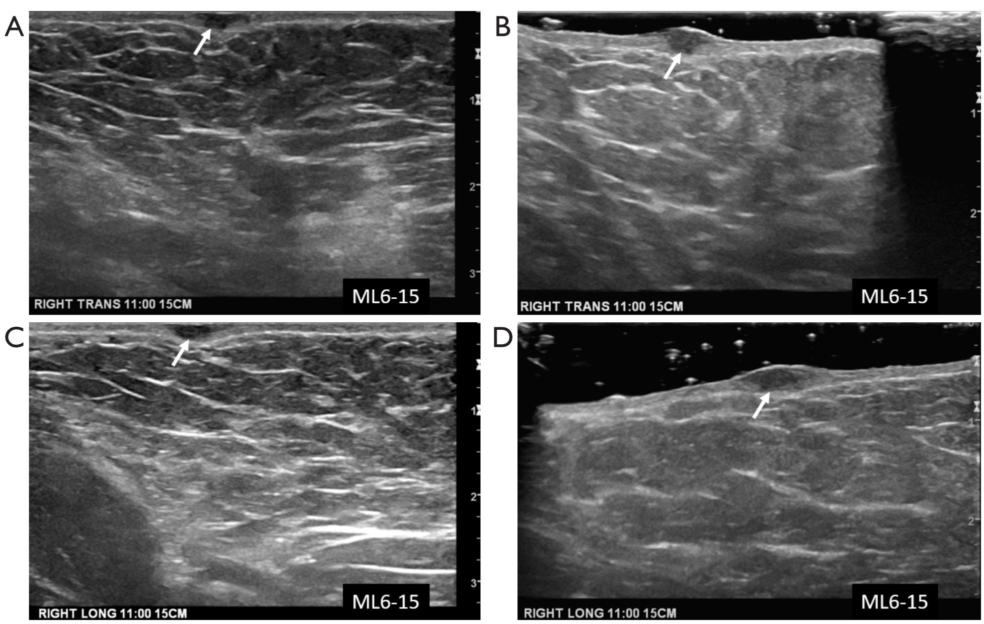


Figure 6 Optimizing focal zones for very superficial findings. This is the same finding shown in *Figure 2*. Using a high-frequency transducer for conventional breast ultrasound, an ML6-15 in this case, a dermal finding (A,C, arrow) is not at the optimal location relative to the focal zones indicated by the hourglass (⧖) symbols on the right-hand side of the image despite setting the focal zones being at their most superficial locations. Using a thicker layer of gel, the dermal finding can be better placed at the focal zones (B,D, arrow).

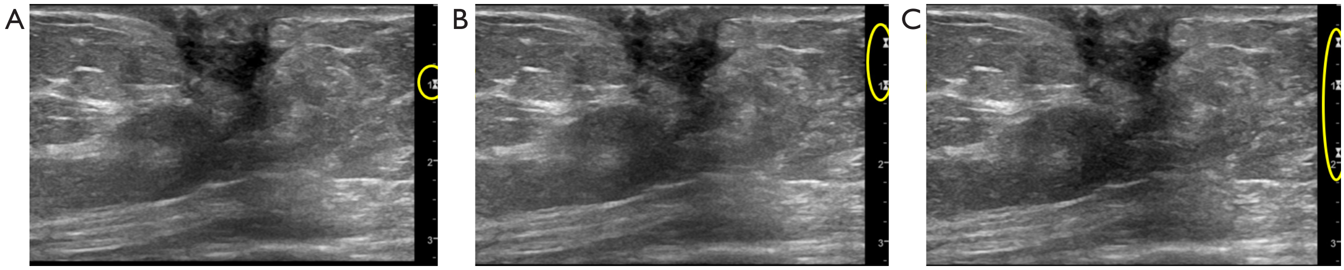


Figure 7 When adjusting focal zones shows no appreciable diagnostic change. In this example, the number of focal zones (A-C, yellow) is increased from 1 (A), to 2 (B), to 3 (C). The (⧖) symbols on the right-hand side of each image (yellow circles) indicate the focal zones. When there is no diagnostic value to increasing the focal zones, one could choose to scan with the fewest focal zones as increasing focal zones decreases the frame rate.

which the energy is deposited in the tissues is called the power [measured in Watts (W) or J/s]. The power or acoustic output knob on an ultrasound scanner is distinct from the on-off button. The power knob is an adjustable knob that controls the acoustic output power of the ultrasound waves which affects the intensity of the sound waves that penetrate the tissues. Increasing the intensity of the sound waves (i.e., the signal) can provide increased penetration of the sound waves, which may be advantageous in some settings, such as when imaging an obese patient (10). However, if the scanned target is associated with reverberations, increasing the power increases the signal arising from these artifacts as well, so the diagnostic value must consider these factors.

High-power deposition of sound waves into tissues can cause bioeffects such as tissue heating and cavitation (11,12). Regulations associated with the power did not arise until 1976 when the United States Food and Drug Administration set a limit for fetal use, but this was later increased in 1992 with the mandate to display the bioeffects of increased power (13). Displayed on ultrasound scanners are the mechanical index (MI) and the thermal index (TI), which change with changes in the power or acoustic output. The MI is unitless with higher values related to the probability of ultrasound waves that can generate very tiny gas bubbles in tissues and cause them to oscillate and break, causing adverse effects. The TI is also unitless and reflects the increase in temperature of the tissues at the focal zone of the ultrasound beam. For example, a TI of 2 would indicate that the temperature of the tissues there could potentially increase by 2 ℃.

Although adverse bioeffects caused by acoustic output in breast ultrasound have not been described (14), one cross-sectional study of various ultrasound exams showed that breast imaging was associated with the highest TI (0.74±0.29) and the lowest MI (0.34±0.54) compared, for example, to thyroid ultrasounds with some of the lowest TI values and ankle ultrasounds with the highest MI values (15). Clinical ultrasound scans should generally be performed with an MI of 0–1.9 and a TI of 0–2.0.

In practice

Awareness of the MI and TI of a scan is particularly important in first-trimester fetal ultrasound when organogenesis occurs, and fetal tissue is most sensitive. Adhering to the principle of as low as reasonably achievable (ALARA) aids in achieving a diagnostic quality exam while maximizing safety. Lowering the power, using color flow Doppler judiciously, and decreasing the scan duration by either using the "Freeze" button when not scanning or avoiding keeping the probe stationary for long periods during scanning can be considered. In practice in breast imaging, the acoustic output is generally at 100% while still maintaining safe MI and TI values (*Figure 8*).

(IV) Depth knob

Breast sonographers are trained to capture most findings in the center of the field of view. The reason is not entirely to frame the finding but to improve frame rates. If the depth of the imaging field of view is 6 cm, sound waves need, at a minimum, to penetrate through at least 6 cm of tissue and return to the transducer. In practice, a little more time is needed to account for non-zero beam angle cases or for the somewhat more theoretical scenario when the signal from the leftmost element travels to the rightmost bottom corner of the field-of-view. A shallower depth means it takes less time for the sound waves to travel through this depth and travel back to the transducer. This increases the frame rate or how quickly the

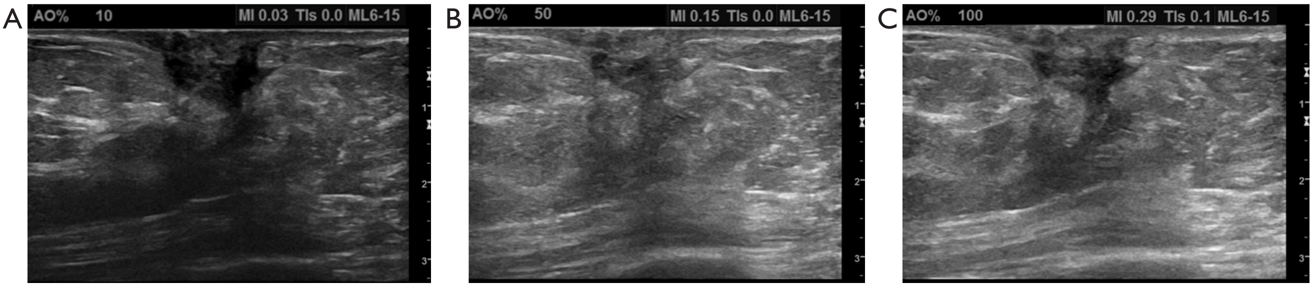


Figure 8 Power (acoustic output) knob. In this example, which is the same finding shown in *Figure 7*, the AO is changed to 10% (A), 50% (B), and 100% (C) while keeping all other scanning parameters constant. Increasing the AO increases the MI and the TI. The default AO is set to 100% (C). AO, acoustic output; MI, mechanical index; TI, thermal index.

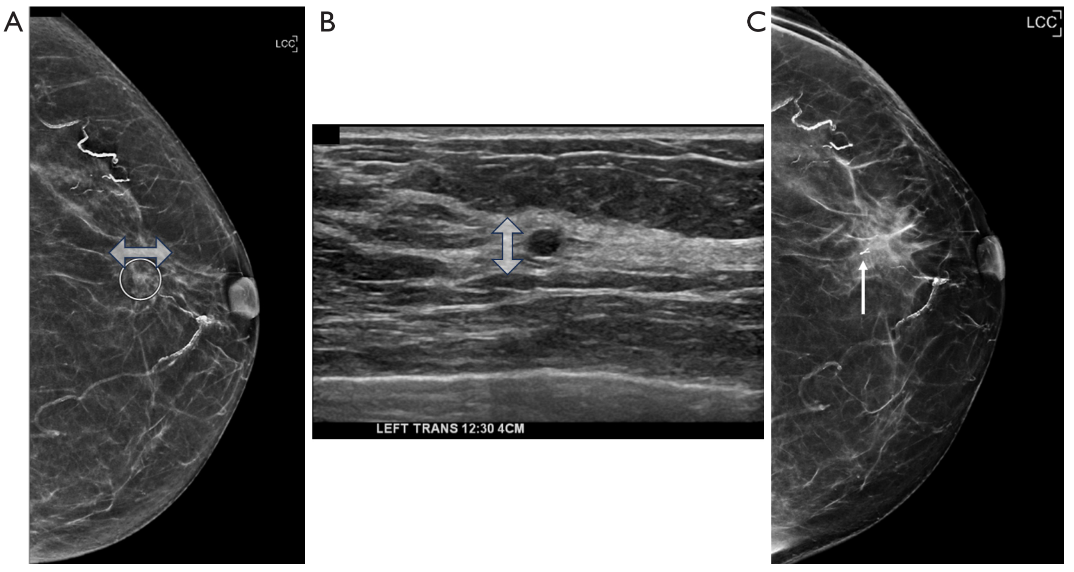


Figure 9 Using depth to frame a finding relative to surrounding anatomy. On the CC view, a mammographic finding (A, circle) is within dense fibroglandular tissue (A, arrow) in the middle-depth breast. Ultrasound of the breast shows a dense band of fibroglandular tissue (B, arrow) that is located middle depth between the skin and the pleura. This finding was biopsied using ultrasound-guidance and post-clip mammogram shows the biopsy marker (C, arrow) corresponding to the expected mammographic location (A) and confirming that what was seen on ultrasound corresponds to the mammographic finding. CC, craniocaudal.

image refreshes; this can be important in evaluating moving structures, such as a pulsing artery. The underlying ultrasound parameter that ties the depth and frame rate together is the PRF, which again refers to the number of ultrasound pulses emitted from the transducer in one second. The higher the PRF, the lower the imaging depth because there is inadequate time for the echoes to travel back to the transducer.

In practice

While it is important to frame a finding so that it is centered in the image, it is also important to acquire at least one image that frames the finding to show the depth of the finding in the breast relative to known anatomic structures. For example, it is helpful to acquire an image of a finding that includes the pectoralis muscle so that it can be determined whether the finding is anterior depth, middle depth, or posterior depth. This can provide more confident correlation with a mammographic finding (*Figure 9*).

(V) Gain and time gain compensation (TGC) knobs

Adjusting the gain of a parameter generally amplifies the

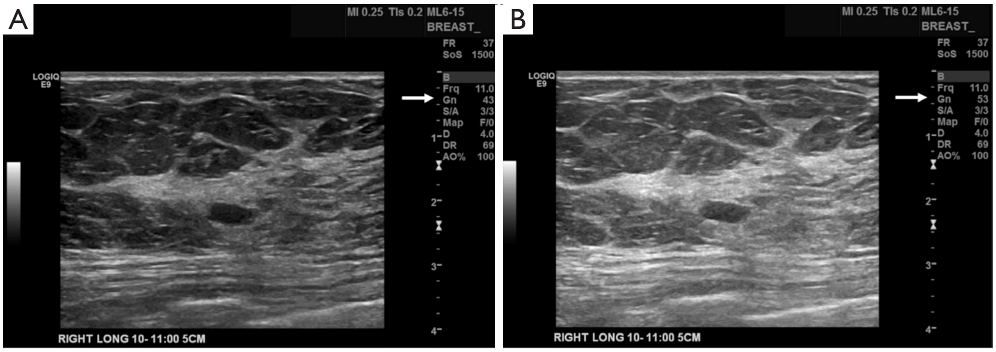


Figure 10 Adjusting the gain. With all the other scanning parameters held constant, adjusting the overall gain (A,B, arrows) was performed to assess the relative echogenicity.

signal associated with that parameter without affecting the MI or TI. In B-mode (brightness or grayscale) ultrasound imaging, increasing the gain increases the overall brightness of the image because increasing the gain amplifies the signals of the echoes after being received by the transducer, effectively making the image brighter. On most ultrasound scanners, there are knobs (typically a collection of sliders on the physical console or touch screen) called TGC or depth gain compensation which allow the operator to adjust the gain at different depths. For instance, when there is a loss in the signal of the echoes due to scattering or absorption (i.e., attenuation) in the deeper tissues, the TGC knobs can be adjusted to preferentially increase the amplitude of the echoes in the deeper tissues returning to the transducer.

In practice

Adjusting the gain and TGC knobs are nearly reflexive for breast sonographers as they assess the echotexture of findings and surrounding tissues. It is important to know that an anechoic finding is not falsely anechoic because the gain was too low (*Figure 10*). When adjusting the gain of the signal, it is important to keep in mind the interrelation between various scanning parameters. For example, changing the B-mode transmit frequency affects the default gain. So, if there is a finding that cannot be optimally depicted, changing the frequency and then the gain or TGC knobs may be helpful.

This is particularly helpful when imaging more challenging anatomy such as the nipple or subareolar region, or when imaging deeper tissues or the axilla.

(VI) Harmonic imaging knob

Sound waves transmitted from the transducer leave the transducer at a fundamental frequency, f_0, that enters the tissues. As the sound waves travel through the tissues, nonlinear effects create harmonic frequencies, which are generally integer multiples of the transmitted fundamental frequency (e.g., $2f_0$, $3f_0$, etc.). Of these harmonic frequencies, the second harmonic frequency, which is two times the fundamental frequency, $2f_0$, provides the most useful diagnostic information (16). The harmonic imaging knob sets the ultrasound scanner so that the harmonic frequencies are processed.

Imaging with harmonic frequencies generally results in fewer artifacts from scatter that arise from the fundamental frequency, particularly in deeper tissues, thereby creating clearer, less noisy images (16). This may be particularly notable in more heterogeneous breast tissues where there are several interfaces between the fatty lobules and fibroglandular tissues that are comprised mostly of suspensory ligaments, glandular components, ducts, and vasculature (*Figure 11*). A trade-off for clearer imaging with harmonic imaging is a slower frame rate because processing the harmonic frequencies takes longer than processing just the fundamental frequency or depending on the vendor may involve transmitting more than one time to create the harmonic image. Since harmonic imaging is often used when imaging deeper structures, the increased imaging depth also contributes to the slower frame rate. But, improving technologies provide strategies to mitigate the slower frame rate. For example, some scanners may use pulse inversion imaging which requires two transmissions for each beam/frame thereby lowering the frame rate by a factor of 2 (17).

In practice

Harmonic imaging can improve the contrast and sharpness of the tissue echotexture that often reflects the heterogeneous nature of breast parenchyma. For example, turning on harmonic imaging can provide another look at the margins

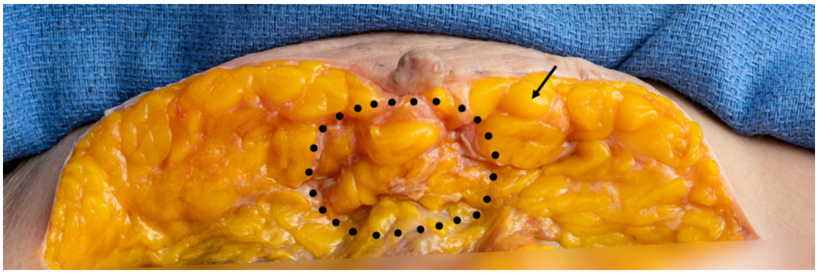


Figure 11 Gross anatomy of breast parenchyma. A longitudinal cross-section of cadaveric breast tissue shows the heterogeneity of the tissues from smooth fatty lobules (arrow) to a region of heterogeneous fibroglandular tissues that include suspensory ligaments (of Cooper), ducts, glandular structures, and vasculature (dotted region). Harmonic imaging can be considered when imaging areas of heterogeneous interfaces.

of masses adjacent to glandular tissue, often improving their conspicuity (*Figure 12*). When using harmonic imaging, the functionality of the gain and TGC knobs also applies. The details of how harmonic imaging is implemented are often vendor-specific, and the operator would need to determine when and how best to use harmonic imaging.

Harmonic imaging can be helpful when searching for axillary lymph nodes (16); this is illustrated in *Figure 13*. While harmonic imaging decreases the frame rate and can be distracting during a procedure, it can still be used during a procedure to provide improved visualization of the target; the slower frame rate could be mitigated by having the sonographer toggle harmonic imaging on and off and if possible decreasing the depth to target. Additionally, harmonic imaging can be used with other transducers (*Figure 14*) or transmit frequencies (*Figure 15*).

With improved spatial resolution and contrast-to-noise ratio, harmonic imaging can be used to "clean up" cystic masses making them more anechoic by reducing any reverberation artifacts. However, it is important to realize that harmonic information alone may not provide sufficient information about the internal architecture of a mass, and additional characterization is needed to distinguish between a solid and a cystic mass.

(VII) Spatial compounding knob

The spatial compounding knob acquires multiple acquisitions of the same area usually at different steering angles to create a composite image. Doing so often increases the signal-to-noise ratio of the composite image improving clarity of details such as margin characteristics (*Figures 16,17*). Spatial compounding also reduces the speckle or graininess of the image.

Speckle occurs when small structures scatter the sound waves causing constructive and destructive interference that creates the speckly pattern. Like harmonic imaging, the details of how spatial compounding is implemented is usually vendor-specific. For example, speckle reduction can involve frame averaging or other filtering algorithms whereby information from more than one frame is combined to improve the signal yielding a less noisy image.

In practice

In breast ultrasound, spatial compound imaging is the default. On occasion, turning off spatial compound imaging may be helpful to evaluate shadowing characteristics of metallic biopsy markers, for example, or to evaluate for posterior acoustic enhancement associated with cysts.

(VIII) Dynamic range

Dynamic range in ultrasound provides control of how one wants to visualize the range of acquired echoes. Dynamic range is similar to window and leveling in computed tomography. Choosing to display a narrow dynamic range results in a high-contrast image, while a wide dynamic range results in a low-contrast image (*Figure 18*).

In practice

Lowering the dynamic range can be used to create a higher contrast image, which may provide more contrast at the margins. However, the trade-off is that some detail in the very high signal areas and in the very low signal areas can be lost. Analogously, a wide dynamic range image results in much less contrast, making it more challenging to distinguish the tissue planes (*Figures 19-21*).

(IX) Beam steering knob

Beam steering allows the operator to angle the beam

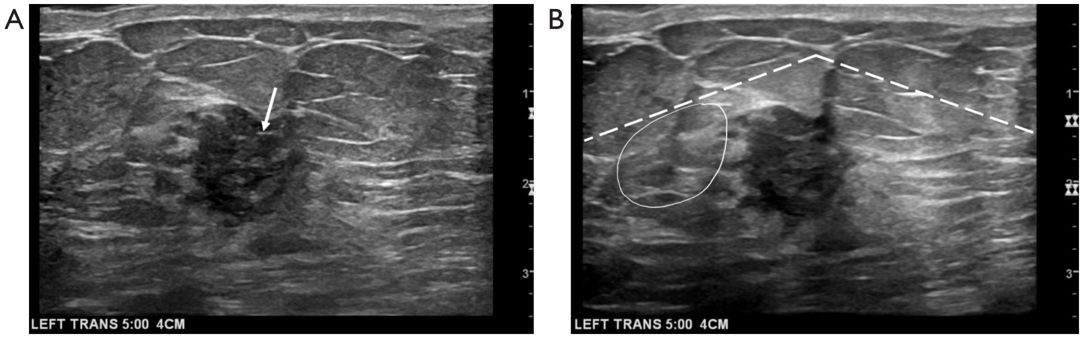


Figure 12 Harmonic imaging for improved sharpness and contrast of parenchymal interfaces. Default conventional breast imaging uses spatial compounding, as seen with the breast cancer in (A, arrow). In (B) harmonic imaging is turned on. Harmonic imaging is referred to as contrast harmonic imaging on the GE Logiq E9 scanner and indicated by the double hourglass icons (⧖⧖) on the right-hand side of the image (B). With harmonic imaging, note the improved sharpness of the interfaces of fatty lobules and the interdigitating fibroglandular tissues (B, circled region). The margins of the mass adjacent to the background breast parenchyma particularly below the white dashed lines in (B) are also sharper.

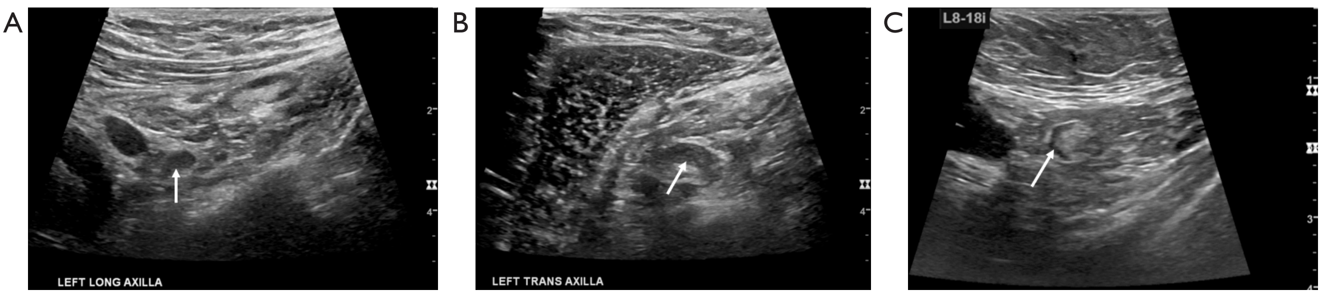


Figure 13 Harmonic imaging for improved sharpness and contrast of axillary lymph nodes. Improved visualization of a deeper axillary lymph node using the ML6-15 transducer is obtained with harmonic imaging in the longitudinal view (A, white arrow) and in the transverse view (B, white arrow). During fine needle aspiration of this lymph node, harmonic imaging can help confirm correct placement of the needle tip. In (C) harmonic imaging can be used with the L8-18i transducer to identify a different axillary lymph node (C, arrow).

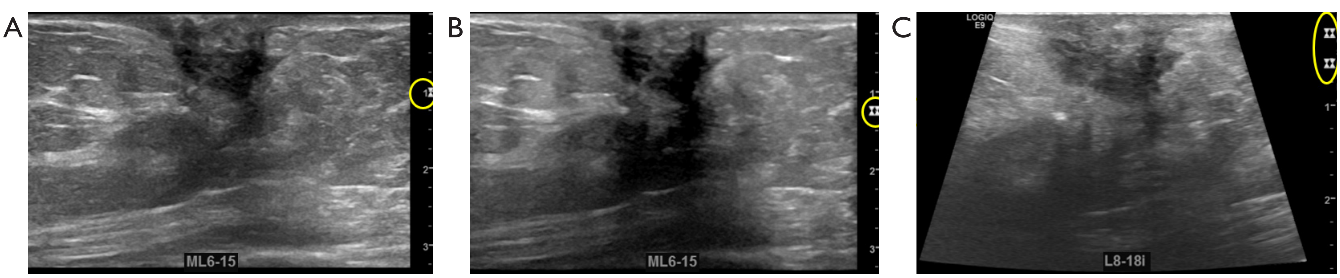


Figure 14 Changing more than one knob. Given this superficial finding (same as the finding in *Figures* 7,*8*) with questionable middle-depth involvement (1–2 cm deep) using the ML6-15 transducer (A), a couple of other techniques were attempted. Harmonic imaging (B) increased the radiologist's confidence in extension of the process to the middle depth, and changing to a higher-frequency transducer (C), the L8-18i, along with harmonic imaging showed persistence of the middle-depth hypoechoic areas which warranted description in the final report.

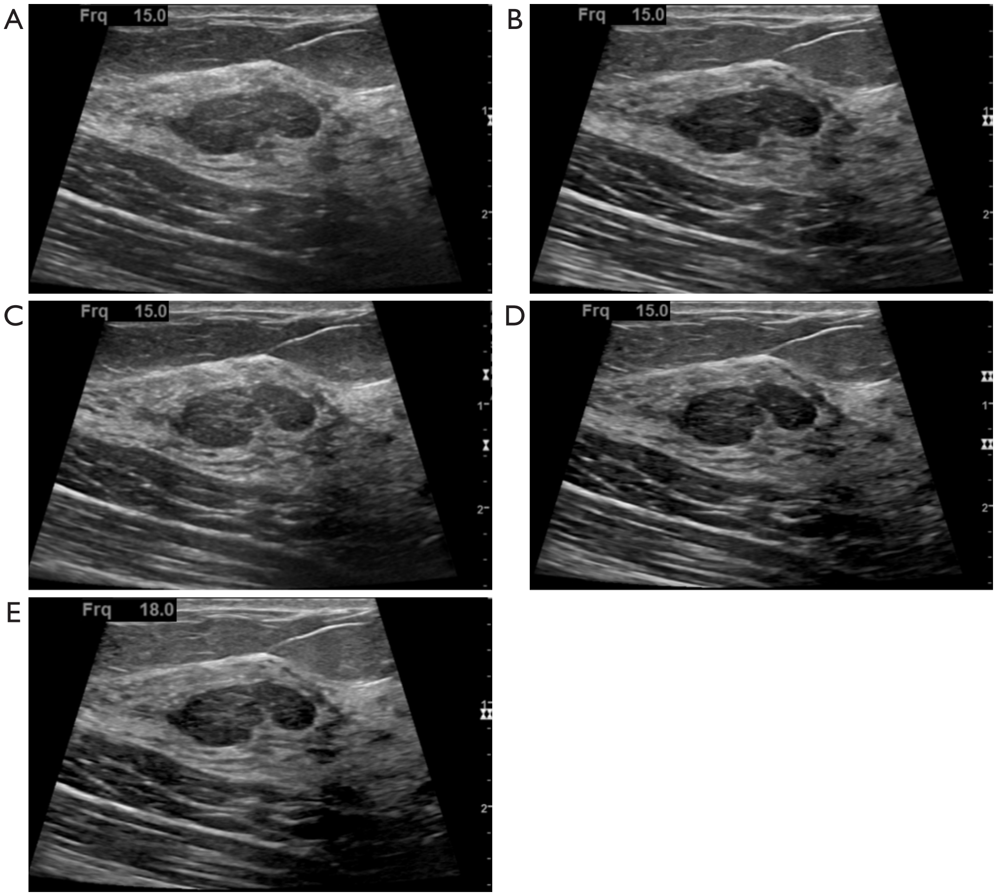


Figure 15 L8-18i transducer at different transmit frequencies with and without harmonic imaging. Realize that one can also change the transmit frequency for a given transducer. Keeping all other scanning parameters the same, harmonic imaging at a transmit frequency of 15 MHz shows slightly better contrast and margin definition (B,D) than the images acquired without harmonic imaging (A,C). In this case, the L8-18i allowed for a transmit frequency of 18 MHz (E) which combined with harmonic imaging shows good conspicuity of the mass.

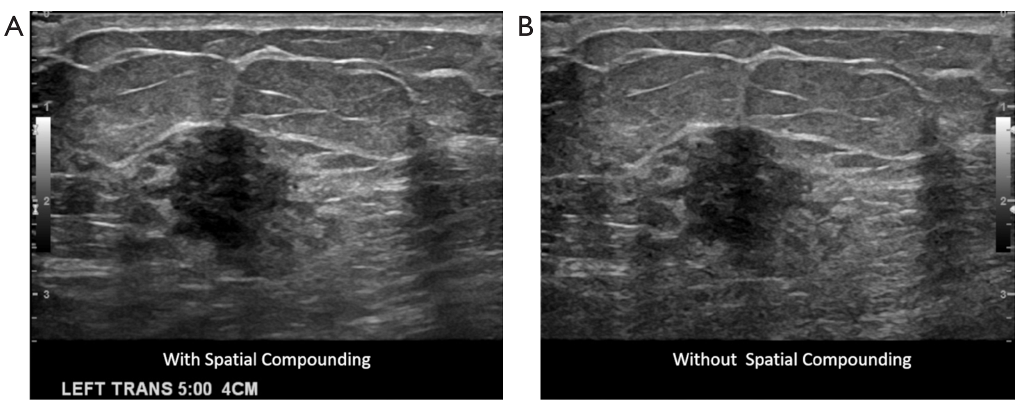


Figure 16 Spatial compounding. By default, spatial compounding is turned on. A breast cancer is shown with (A) and without (B) spatial compounding. When spatial compounding is off, the breast parenchyma surrounding the breast cancer, particularly posteriorly, is slightly noisier, a reflection of single image acquisition. Note that the shadowing features without spatial compounding (B) show slightly more contrast to the adjacent tissues when compared to an image acquisition with spatial compounding (A).

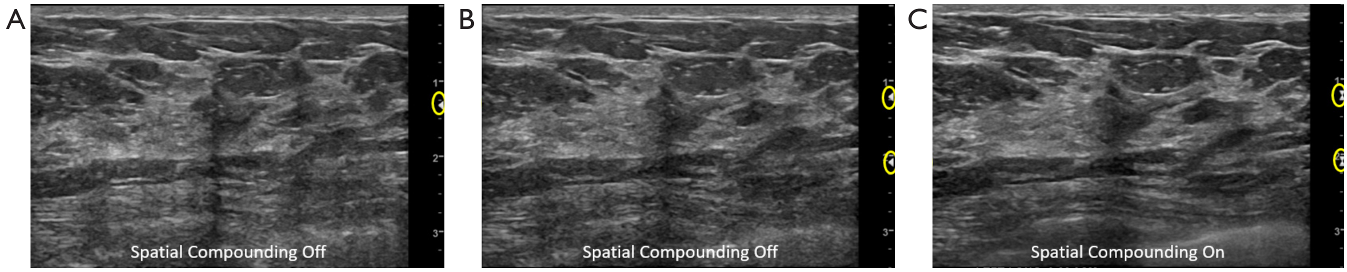


Figure 17 Focal zones and spatial compounding. The irregular hypoechoic mass in the center of each image illustrates slight variations with one focal zone (A) versus two focal zones (B,C). A smoother or softer look is appreciated when spatial compounding in turned on (C) as it involves some speckle reduction and frame averaging which is vendor-specific.

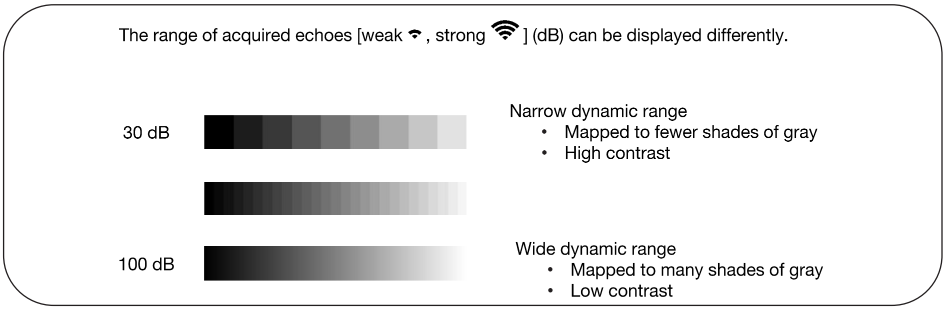


Figure 18 Dynamic range. In this illustration, a narrow dynamic range, such as 30 dB, maps the acquired echoes to fewer shades of gray, increasing the contrast of the image. Similarly, a wider dynamic range, such as 100 dB, maps the acquired echoes to more shades of gray, essentially reducing the contrast of the image. This can be appreciated in the horizontal gray bars, where the contrast between the shades of gray is more readily apparent with a narrow dynamic range (30 dB) compared to a wider dynamic range (100 dB).

Figure 19 DR and background echotexture. Holding all other scanning parameters constant, the dynamic range has been varied from the narrowest DR of 36 dB in (A), to the default DR of 69 dB (B), and then to the widest DR of 96 dB (C). The higher contrast is evident with the narrowest DR in (A). Note that in high contrast images (A), the details can be lost in very high signal areas and in very low signal areas. In comparison, very low contrast images (C) can effectively efface heterogeneous echotexture of the background breast parenchyma. DR, dynamic range.

Figure 20 DR and margins. Holding all other scanning parameters constant, the DR has been varied from the narrowest DR of 36 dB in (A), to the default DR of 69 dB (B), and then to the widest DR of 96 dB (C). The high-contrast image with a DR of 36 dB (A) provides some detail of the margins of the mass not well appreciated when the DR is increased (C). Again, with low DR, details are lost in very high signaling areas (A, white areas) and in very low signaling areas (A, black areas). DR, dynamic range.

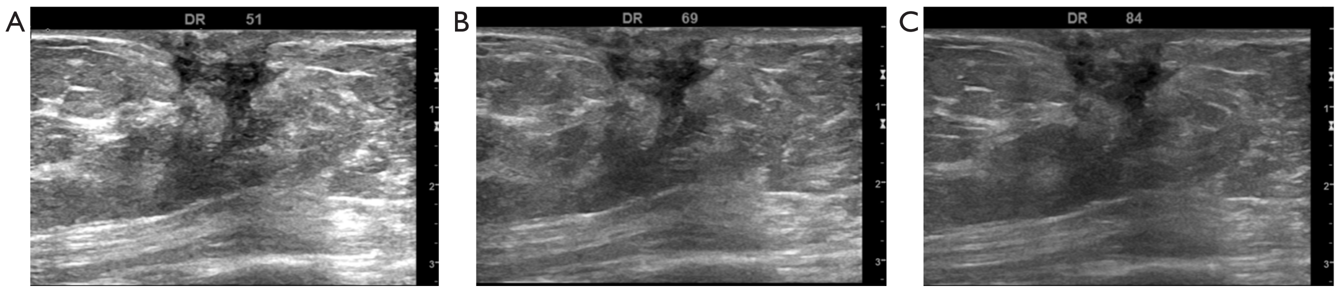


Figure 21 DR and visualization preferences. The irregular hypoechoic mass shown is the same one in *Figures 7,8,14*. Holding all other scanning parameters constant, the DR has been varied from a DR of 51 dB in (A), to the default DR of 69 dB (B), and then to a DR of 84 dB (C). Depending on what feature one is focusing on, one can vary the DR knob to provide more contrast to certain areas by lowering the DR. DR, dynamic range.

of ultrasound waves emitted from the transducer. This effectively directs the beam of waves so their angle of insonation in tissues is changed, potentially improving the visualization.

In practice

Beam steering can be particularly helpful when looking for the needle tip during a fine-needle aspiration (FNA) of a deep axillary lymph node (*Figure 22*, Videos S1,S2).

(X) Color Doppler knob for twinkling detection of breast biopsy markers

Color Doppler is an ultrasound technique that can determine the velocity of moving structures, usually blood flow. Color Doppler twinkling, on the other hand, is characterized by a temporally changing color mosaic pattern that was first described with kidney stones in 1996 (18). Literature over the past three years reports ultrasound color Doppler twinkling as a readily available ultrasound technique for detecting some breast biopsy markers such as the Tumark Q (Hologic, Marlborough, MA, USA), the Trimark Cork (Hologic), and the Tumark MRI-safe Flex (Hologic) markers (6-8,19). The literature has also described some twinkling associated with the Twirl Ring marker (20).

Twinkling-detection of breast biopsy markers is currently reduced to five steps related to the probe (transducer), color Doppler frequency, gain, and scale.

Probe: select a lower-frequency range transducer (*Figure 23*), such as a curvilinear transducer with a transmit frequency of 1–6 MHz, which is often used in abdominal ultrasound imaging.

Color flow Doppler: color flow Doppler is a knob that is available on nearly every ultrasound scanner and is generally used for quantifying blood flow (*Figure 24*) with scanning techniques that often consider compression and the angle of the beam. But for twinkling detection, the color Doppler knob simply needs to be turned on without too much consideration paid to operator-dependent scanning. While the underlying cause of twinkling is still under

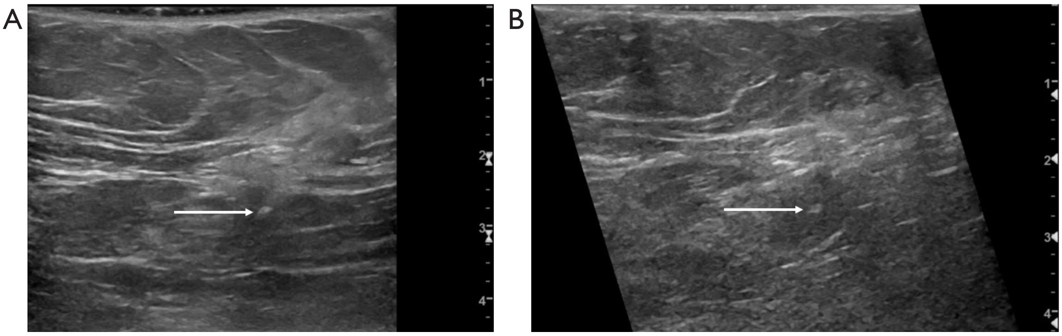


Figure 22 Steering. These two static images show fine-needle aspiration of an axillary lymph node without (A) and with (B) steering. Steering is evident by the angled field of view (B). For this particular ultrasound vendor, steering requires disabling spatial compounding, so the steered image (B) appears less smooth. Without steering, the default spatial compounding results in a smoother image appearance (B). While the needle tip (A,B, arrow) can be seen without (A) and with (B) beam steering, preference for steering is user dependent. The performance of steering is sometimes better appreciated in cine clips. Two cine clips without and with steering (Videos S1,S2) are provided in the supplemental section.

Probe
- GE Loqiq E9 & E10

Parameter	ML6-15	9L	C1-6
B-mode for Breast	Excellent	Mid-range	Poor
Twinkling	Poor	Mid-range	Excellent
Depth	Superficial	Mid-range	Deep (Abdominal)
Color Doppler Frequencies	6.3, 7.5, 8.3, 10.0, 11.9,12.5 MHz	3.1, 3.6, 4.2, 5.0, 6.3 MHz	1.7, 1.9, 2.1, 2.5, 3.1, 3.6 MHz

- Transducer names imply frequency bandwidth (range)
- ACR Breast Ultrasound Requirements: ML6-15

Figure 23 Twinkling and transducer selection. While color Doppler twinkling is more readily detected using a lower frequency transducer, the anatomic detail in the breast using this transducer will be poor. Immediate subsequent interrogation of the site of twinkling using a higher frequency transducer typically used in breast ultrasound is then performed to identify features of the breast biopsy marker which can be subtle (8). ACR, American College of Radiology.

active investigation, twinkling is a compelling qualitative diagnostic feature that is readily recognized.

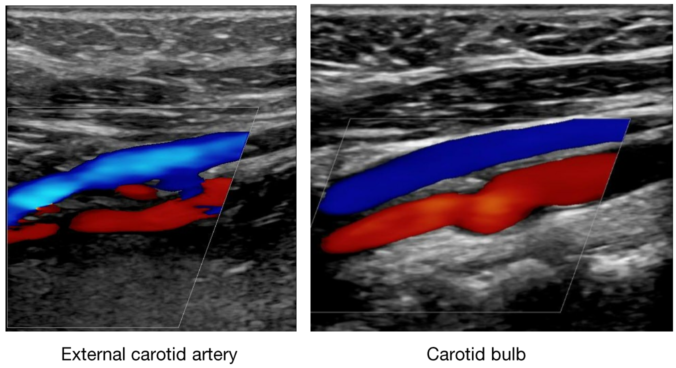
Frequency: after color Doppler is turned on, there is an option for selecting the color Doppler transmit frequency (*Figure 25*). Selecting a color Doppler transmit frequency (f_0) will change the scale (cm/s) because of the relationship between color Doppler transmit frequency (f_0) and velocity (v) for a given Doppler shift (f_d). Color Doppler twinkling changes with color Doppler frequencies, and lower color Doppler transmit frequencies typically lead to better twinkling.

Gain: as described before, gain affects the amplification of the overall signal measured. In this case, it is the Doppler gain, and adjusting the Doppler gain affects the strength or amplitude of the ultrasound signal affecting the appearance of the twinkling (*Figure 26*). Typical starting values are 60–80% of the maximum. High gain values may introduce noise from hand motion or patient respiration. Low gain values may suppress twinkling signals.

Scale: the scale associated with color Doppler adjusts

Color flow doppler

- Most often used for blood flow quantification
- B-mode transmit frequency
  - Determines frequency used for grayscale image
- Color doppler transmit frequency
  - Used for flow velocity evaluation



$$f_d = \frac{2 f_0 v \cos\theta}{c}$$

$$v = \frac{f_d c}{2 f_0 \cos\theta}$$

f_d: Doppler shift, Hz
f_0: Doppler transmit frequency, MHz
v: Flow velocity, m/s (cm/s)
c : Sound speed (1,540 m/s)
θ: Angle between US beam and vessel

Figure 24 Twinkling and color flow Doppler. Color Doppler transmit frequency is different than B-mode transmit frequency but is dependent on the B-mode transmit frequency, which is inherent with the transducer selected. For twinkling detection, the operator will need to select a more favorable f_0, which is later described. US, ultrasound.

Frequency

- Frequency is selectable (red box) and will change the scale due to the relationship between transmit frequency and velocity for a given doppler shift.
  - This can be changed to see if twinkling improves.
  - Typically, lower transmit frequencies lead to better twinkling.

Parameter	ML6-15	9L	C1-6
Color Doppler Frequencies	6.3, 7.5, 8.3, 10.0, 11.9,12.5 MHz	3.1, 3.6, 4.2, 5.0, 6.3 MHz	1.7, 1.9, 2.1, 2.5, 3.1, 3.6 MHz

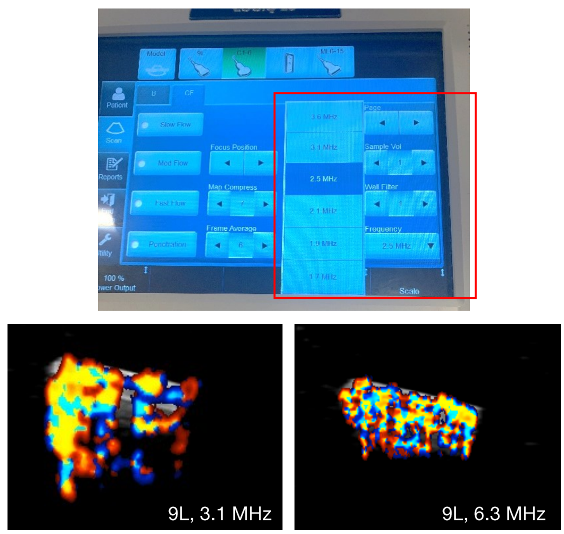


Figure 25 Twinkling and color Doppler transmit frequency. Twinkling detection is more favorable for lower frequency transducers such as the 9L and the C1–6 transducers, and for each of these transducers, the lower color Doppler transmit frequencies are favored. For the 9L transducer, using a color Doppler transmit frequency ranging 3.1–4.2 MHz is favored; and for the C1–6 transducer, using a color Doppler transmit frequency ranging 1.7–3.1 MHz is favored. One of the characteristics of true twinkling is its insensitivity to changes in color Doppler transmit frequencies (7).

the range of velocities measured by the Doppler imaging (*Figure 27*). It is usually a range of velocities between –*v* cm/s to +*v* cm/s. So, when imaging an artery, the scale can be increased to reflect higher blood flows in the arteries compared to the veins, and also to minimize aliasing. In the context of using color Doppler for detection of twinkling, adjusting the scale can be used to eliminate vascular flow, particularly when trying to minimize any false-positive

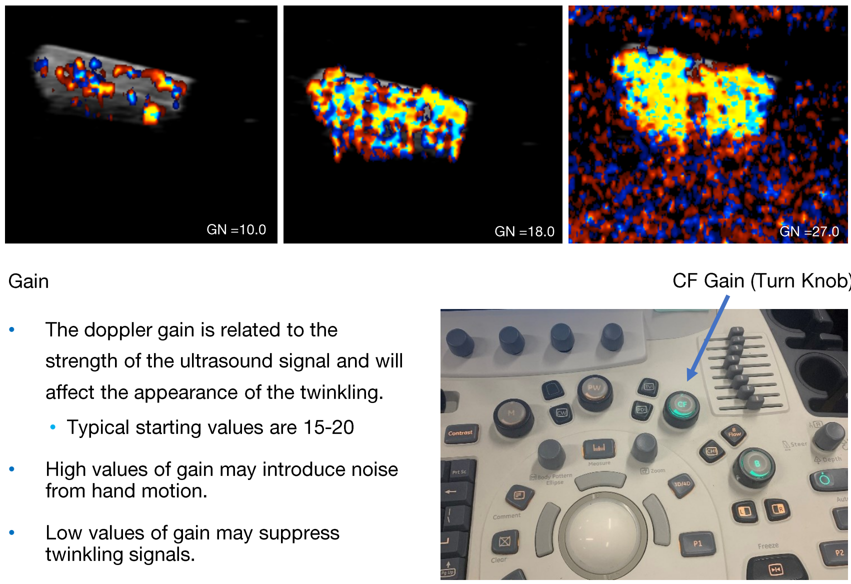


Figure 26 Twinkling and Doppler gain. Doppler GN is operator-dependent and strives to strike a balance between twinkling and noise. The maximum gain on this ultrasound scanner is 30.0. When the GN is set too low (GN =10.0), noise is filtered out but so is a substantial component of twinkling. When the GN is set too high (GN =27.0), noise can confound the twinkling signature. The GN is often selected to just below the level of noise, in this case GN =18.0, that demonstrates twinkling with very minimal, if any, noise. GN, gain; CF, color flow.

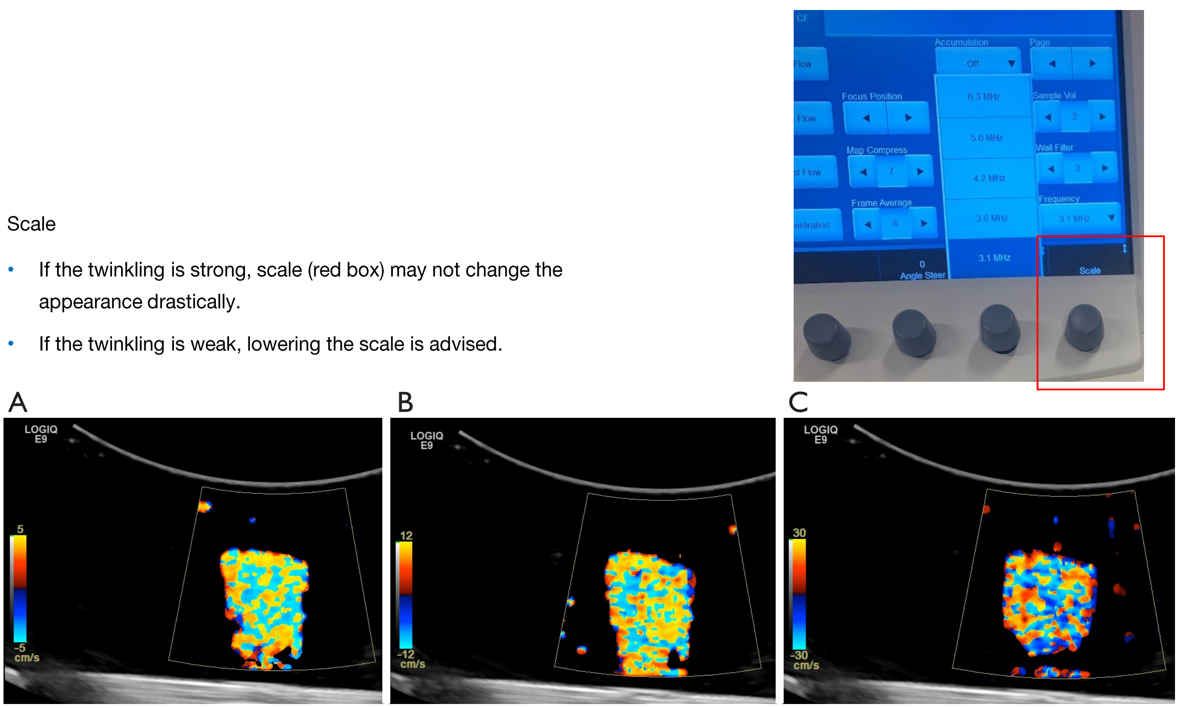


Figure 27 Twinkling and color Doppler scale. Twinkling in (A-C) is robust despite changes to the scale ±5 cm/s (A), ±12 cm/s (B), and ±30 cm/s (C).

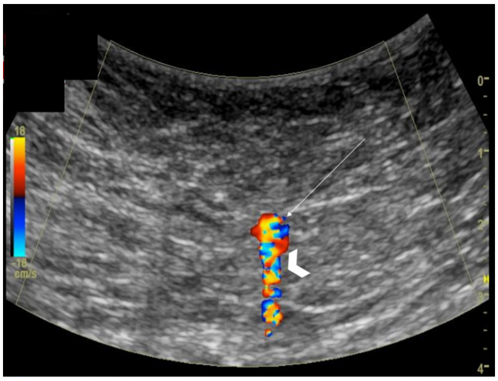


Figure 28 Color Doppler ultrasound twinkling-detection of markers. Using a curvilinear lower frequency transducer, the C1–6, color Doppler twinkling readily characterizes this biopsy marker, in this case a Tumark Q marker (arrow). Earlier extensive B-mode scanning with the conventional ML6-15 transducer could not identify the marker. Note that the twinkling comet tail (chevron) is also seen.

twinkling that is arising from vascular flow. If the twinkling is weak, lowering the scale may be helpful. If the twinkling is strong, adjusting the scale may not appreciably change the appearance of the twinkling.

In practice

Ultrasound color Doppler twinkling is readily apparent for some commercial breast biopsy markers (Tumark Q, TriMark Cork, and Tumark MRI-safe Flex) (6-8,19) and for markers made from polymethyl methacrylate (*Figure 28*) (21).

Color Doppler is vendor agnostic making twinkling of markers a relatively straightforward scanning technique by adjusting four knobs: the color Doppler knob (turn it on); the color Doppler frequency; the gain; and the scale. This technique is summarized in Appendix 1 that summarizes tips for using color Doppler twinkling to detect some breast biopsy markers. Those who like to use mnemonics can consider "Penguins Can't Fly Gracefully South" to recall the roles of probe (transducer) selection, color Doppler, color Doppler frequency, gain, and scale.

Discussion

Ultrasound is integral to any breast imaging practice, and awareness of the functionalities of the knobs on the ultrasound console is critical to ensure that imaging features accurately characterize a finding. Like other imaging modalities, ultrasound involves trade-offs. It is important to recognize that optimizing scanning parameters to enhance image quality will compromise other parameters, so being able to discern diagnostic features from artifacts is essential. The ever-questioning breast sonographer will always ask whether the scanning parameters have inadvertently made cancer look like a benign cyst which usually leads to unfortunate consequences. Skills in breast ultrasound must be dynamic and continuously improved to keep pace with advancing ultrasound technologies and to troubleshoot challenging patient cases effectively. This manuscript provides a vendor-agnostic, high-level review of ten frequently used knobs in breast ultrasound along with practical tips relevant to each knob.

Many ultrasound scanners offer an automated feature that optimizes scanning parameters without the user knowing exactly what may have been altered. Knowing how to optimize an ultrasound scan without the automated feature gives one more control over what one sees. Oftentimes, ultrasound image acquisitions are based on what the practice or individual breast radiologist prefers. Because of the varied preferences, many breast radiologists still back-scan after the breast sonographer, not so much to rectify a scanning error, but to better appreciate the imaging findings in real-time.

Many breast ultrasound practices establish scanning protocols for various clinical indications. For instance, a protocol for axillary imaging may include how many transverse and longitudinal images to take and how many images to take with measurements. Such protocols can add uniformity to scans and are usually executed after practice consensus. "Recipes" for how to approach a breast ultrasound may include tips such as "Use Auto Optimize to add contrast to your image as needed". While such recipes may be helpful for those onboarding, de facto use of knobs has the potential to miss opportunities to optimize scanning parameters and acquire higher-quality, more convincing diagnostic ultrasound images. With advancing technologies in ultrasound scanners that tout nuanced differences based on vendors, basic breast ultrasound knobology remains the same, and the quality of breast ultrasound will still be very much operator-dependent. Therefore, a better understanding of breast ultrasound knobology can benefit the sonographer, the breast radiologist, and most importantly, the patient.

Ultrasound color Doppler twinkling for detecting some breast biopsy markers is a readily available technique that is relatively easy to execute. Of the more than three dozen breast biopsy markers commercially available, not all markers twinkle, but those that do twinkle, such

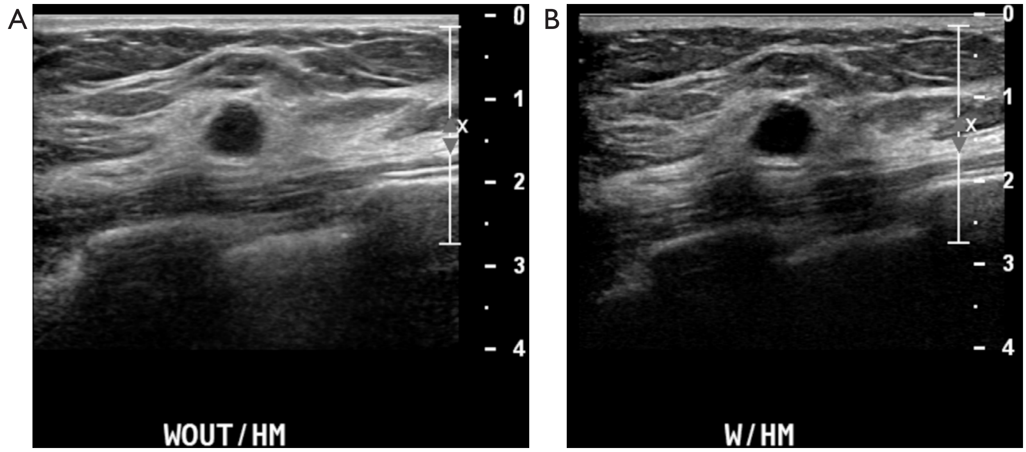


Figure 29 Comparing knobs to maximize characterization of features. The indistinct margins without harmonic imaging (A) persist with harmonic imaging (B). The hypoechoic internal echogenicity of the mass without harmonic imaging (A) outweighs the harmonic images with through-transmission and apparent anechoic internal characteristics that might suggest a cyst or cystic component. This mass was biopsied and was an invasive ductal carcinoma.

as the Tumark Q, the Trimark Cork, the Tumark MRI-safe Flex, and a marker made of polymethyl methacrylate (21,22), improved confidence in identifying these markers for ultrasound-guided preoperative localization can make a significant impact on the management of patients with breast cancer.

One size does not fit all in breast ultrasound. More often than not, the ultrasound target is indeterminate, and ultrasound has to provide the best representation of relevant features of the finding that will ultimately determine the level of suspicion and the management. For example, distinguishing a benign cyst that may be slightly inflamed or contain internal debris from a cancer is often non-trivial. In *Figure 29*, the sonographer took a side-by-side image of a finding without and with harmonics. The information provided by acquiring images with and without harmonic imaging is non-contradictory when one realizes what harmonic imaging is acquiring. Assimilating nuanced information from additional image acquisitions can help with the interpretation.

Conclusions

A breast radiologist is only as good as the quality of the images acquired. Breast ultrasound knobology with additional consideration of ultrasound twinkling for marker detection, as presented here, hopes to help and empower those who perform breast ultrasounds to acquire the images that can make a positive difference in patient care.

Acknowledgments

Funding: The effort for this manuscript was funded in part by the Mayo Clinic Discovery Translation Program fund. This work was also funded in part by grant R01 EB033008 from the National Institute of Biomedical Imaging and Bioengineering (NIBIB). The content is solely the responsibility of the authors and does not necessarily represent the official views of the NIBIB or the National Institutes of Health (NIH).

Footnote

Peer Review File: Available at https://tbcr.amegroups.org/article/view/10.21037/tbcr-24-30/prf

Conflicts of Interest: All authors have completed the ICMJE uniform disclosure form (available at https://tbcr.amegroups.org/article/view/10.21037/tbcr-24-30/coif). C.U.L. serves as an unpaid editorial board member of *Translational Breast Cancer Research* from October 2023 to September 2025. C.U.L. and M.W.U. are co-PIs on a National Institute of Biomedical Imaging and Bioengineering (NIBIB) grant (R01 EB033008) to the Mayo Clinic and on an internal Mayo Clinic Discovery Translation Program award. Patents are pending for non-metallic ultrasound detectable markers as well as Doppler ultrasound twinkling technologies. G.K.H. was supported by NIBIB grant (R01 EB033008) to the Mayo Clinic and by an internal Mayo Clinic Discovery Translation Program award. The other authors have no other conflicts of interest to declare.

Ethical Statement: The authors are accountable for all aspects of the work in ensuring that questions related to the accuracy or integrity of any part of the work are appropriately investigated and resolved.

References

1. Nicosia L, Gnocchi G, Gorini I, et al. History of Mammography: Analysis of Breast Imaging Diagnostic Achievements over the Last Century. Healthcare (Basel) 2023;11:1596.
2. WILD JJ, NEAL D. Use of high-frequency ultrasonic waves for detecting changes of texture in living tissues. Lancet 1951;1:655-7.
3. Fornage BD, Toubas O, Morel M. Clinical, mammographic, and sonographic determination of preoperative breast cancer size. Cancer 1987;60:765-71.
4. Lee J. Practical and illustrated summary of updated BI-RADS for ultrasonography. Ultrasonography 2017;36:71-81.
5. Lee CS, Bohm-Velez M. ACR Practice Parameter for the Performance of a Diagnostic Breast Ultrasound Examination. Am Coll Radiol 2021. Available online: https://www.acr.org/-/media/ACR/Files/Practice-Parameters/US-Breast.pdf
6. Jakub JW, Hesley GK, Larson NB, et al. Ultrasonographic Detection and Surgical Retrieval of a Nonmetallic Twinkle Marker in Breast Cancer: Pilot Study. Radiol Imaging Cancer 2022;4:e220053.
7. Lee CU, Larson NB, Urban MW, et al. Factors Associated With Ultrasound Color Doppler Twinkling by Breast Biopsy Markers: In Vitro and Ex Vivo Evaluation of 35 Commercially Available Markers. AJR Am J Roentgenol 2023;220:358-70.
8. Lee CU, Piltin MA, Moldoveanu D, et al. Using US Twinkling Artifact to Identify Breast Biopsy Markers: Brief Report. Radiol Imaging Cancer 2023;5:e220168.
9. Hangiandreou NJ. AAPM/RSNA physics tutorial for residents. Topics in US: B-mode US: basic concepts and new technology. Radiographics 2003;23:1019-33.
10. Cibull SL, Harris GR, Nell DM. Trends in diagnostic ultrasound acoustic output from data reported to the US Food and Drug Administration for device indications that include fetal applications. J Ultrasound Med 2013;32:1921-32.
11. Quarato CMI, Lacedonia D, Salvemini M, et al. A Review on Biological Effects of Ultrasounds: Key Messages for Clinicians. Diagnostics (Basel) 2023;13:855.
12. Fowlkes JB; Bioeffects Committee of the American Institute of Ultrasound in Medicine. American Institute of Ultrasound in Medicine consensus report on potential bioeffects of diagnostic ultrasound: executive summary. J Ultrasound Med 2008;27:503-15.
13. Sheiner E, Freeman J, Abramowicz JS. Acoustic output as measured by mechanical and thermal indices during routine obstetric ultrasound examinations. J Ultrasound Med 2005;24:1665-70.
14. Moyano DB, Paraiso DA, González-Lezcano RA. Possible Effects on Health of Ultrasound Exposure, Risk Factors in the Work Environment and Occupational Safety Review. Healthcare (Basel) 2022;10:423.
15. Fatahi Asl J, Farzanegan Z, Tahmasbi M, et al. Evaluation of the Scan Duration and Mechanical and Thermal Indices Applied for the Diagnostic Ultrasound Examinations. J Ultrasound Med 2021;40:1839-50.
16. Anvari A, Forsberg F, Samir AE. A Primer on the Physical Principles of Tissue Harmonic Imaging. Radiographics 2015;35:1955-64.
17. Simpson DH, Chin CT, Burns PN. Pulse inversion Doppler: a new method for detecting nonlinear echoes from microbubble contrast agents. IEEE Trans Ultrason Ferroelectr Freq Control 1999;46:372-82.
18. Rahmouni A, Bargoin R, Herment A, et al. Color Doppler twinkling artifact in hyperechoic regions. Radiology 1996;199:269-71.
19. Lee CU, Hesley GK, Uthamaraj S, et al. Using Ultrasound Color Doppler Twinkling to Identify Biopsy Markers in the Breast and Axilla. Ultrasound Med Biol 2021;47:3122-34.
20. Tan MP, Bi Z, Ong EMW. The 'twinkle' artifact - A novel

method of clip identification to facilitate targeted axillary surgery following neoadjuvant chemotherapy in breast cancer patients. Clin Imaging 2020;68:36-44.
21. Lee CU, Urban MW, Lee Miller A 2nd, et al. Twinkling-guided ultrasound detection of polymethyl methacrylate as a potential breast biopsy marker: a comparative investigation. Eur Radiol Exp 2022;6:26.
22. Lee CU, Urban MW, Hesley GK, et al. Long-Term Ultrasound Twinkling Detectability and Safety of a Polymethyl Methacrylate Soft Tissue Marker Compared to Conventional Breast Biopsy Markers-A Preclinical Study in a Porcine Model. Ultrasound Med Biol 2024;50:1001-9.

doi: 10.21037/tbcr-24-30
Cite this article as: Lee CU, Hesley GK, Pierson TA, Higgins RL, Urban MW. Breast ultrasound knobology and the knobology of twinkling for marker detection. Transl Breast Cancer Res 2024;5:28.

