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Abstract. In B-mode imaging of the dependent or compressed breast, wave incidence at steep angles can change propagation directions and induce areas of signal dropout. To evaluate the image anomalies in reasonable simulation times, we performed full-wave studies for center frequencies of 1 and 4 MHz. Speed of sound and density of skin, typical coupling gel, and adipose tissue were assigned to the test couplant. Compared with commercial gel, skin-like couplant reduced the dropout area at 1 and 4 MHz by 57.1% and 96.7%, respectively, consistent with a decreased average beam deflection in the breast. Conversely, the adipose-like couplant increased the dropout area from that of simulated commercial gel by 26.5% and 36.7% at 1 and 4 MHz, respectively. In addition, the skin-like couplant resulted in the greatest beam deflection inside the breast among all couplants. The findings could aid the use of three-dimensional simulations to design ultrasound couplants for beam passage through tissue boundaries at steep angles to improve corrections of signal dropout and defocusing and in compound imaging. © 2017 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.4.4.045001](https://doi.org/10.1117/1.JMI.4.4.045001)]

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1 Introduction

In 2016, ~246,660 women in the United States were diagnosed to have breast cancer. Furthermore, out of all estimated new cancer cases, breast cancer accounts for 29%.¹ In fact, breast cancer is the most frequently diagnosed cancer in women. In the past decade, there has been great demand for ultrasound in the United States for breast cancer screening as a supplement of x-ray mammography, since ultrasound can detect unsuspected, mammographically occult cancer, especially in radiographically dense breasts that have a higher risk of developing breast cancer.^{2–4} Also, several tissue acoustical properties obtained from received ultrasound signals such as speed of sound,^{5–13} attenuation,^{8–10,12,14–17} density,¹⁸ and elasticity^{19–21} can be used for characterizing different tissue types as they tend to have different acoustic properties. The ability of ultrasonic imaging to characterize tissues, particularly the differentiation of malignant lesions from benign tissues, shows promise in decreasing the number of breast biopsies needed for adequate diagnosis.²²

Compared with x-ray, ultrasound beams undergo much more complex interactions with the tissues through which they propagate. In some cases, this interaction can alter the propagation patterns substantially.^{23–28} Several studies have tried to extract tissue acoustic properties based on such interactions. To date, however, full-wave calculations have not been performed to study how the imaging anomalies at oblique incidence to the breast explored here arise and can be modified by the choice of coupling agents. Recently, it has been observed that in breast imaging, where images are acquired in the CC-view or in the

mammographic view, the anomalies occur as parts of areas that have limited insonification, i.e., signal dropout areas, and are located close to the papilla and the curved skin layer, where many malignant structures are found.^{29–31} Figure 1 shows the signal dropout areas presented in three different B-mode images, acquired from three different patients using our mammography-configured automated ultrasound scanner developed in-house.^{32–34} As can be seen, signal dropout areas occurring around the papilla could negatively result in misinterpretations in breast cancer diagnosis and substantially increasing further procedures for definitive diagnosis. Considering this, it is of particular interest that automated ultrasonic imaging of the breast in the mammographic geometry, where the ultrasound signal goes through couplant and penetrates the skin at an angle (non-perpendicular direction), oftentimes misses much information of the subareolar tissues due to bending of the acoustic beam by refraction and reflection, leading to the low signal-to-noise ratio (SNR).

Here, we conduct finite-difference-time-domain (FDTD) simulations to investigate artifacts of ultrasound wave propagation in a simulated breast geometry with a curved skin layer under couplants with different acoustic impedances: commercial-gel-like, skin-like, and adipose-like. In addition, we investigate the characteristics of anomalies in the image as results of using the aforementioned couplants, as well as compare the imaging performance of each couplant. The findings of our study will provide detailed information about complex ultrasound signal interactions that occur during breast imaging, which could potentially be helpful for designing a couplant that minimizes these artifacts.

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Fig. 1 Clinical images of compressed female breasts acquired using a GE Logiq 9 ultrasound scanner (GE Healthcare, Milwaukee, Wisconsin) at 10 MHz. Each signal dropout area is enclosed by a manually segmented blue contour. Note that the imaging array is located at the bottom of the images.

This paper is organized as follows: Sec. 2 describes the governing ultrasound wave equations, how the simulation environments were setup, and how the simulation datasets were acquired and processed. Section 3 demonstrates the characteristics of the anomalies resulting from using different couplants and provides an in-depth analysis of ultrasound wave interactions/propagation patterns under different circumstances. Finally, Sec. 4 provides a summary and possible future directions of our study.

2 Materials and Methods

2.1 Breast Geometry

The simulation breast phantom used for all studies consists of ultrasound couplant, skin, and adipose tissue, with their acoustical properties given in Table 1 and their geometry diagrammed in Fig. 2. The dimensions of the field of view in the axial and lateral directions are 37.5 and 25 mm, respectively. The center of the field of view is located at (lateral, axial) = (0, 21.25) mm. The skin layer with a thickness of 2 mm^{35,36} corresponds to the enclosed boundaries that are defined as two elliptical arcs cocentered at $(-12.5 - \Delta x, 37.5)$ mm, where Δx is the lateral shift of the transducer, and with half major axes of 31 and 33 mm and half minor axes of 18 and 20 mm. By means of Δx , lateral beams are simulated by varying Δx from 0 to 8 mm. The bottom of the numerical breast is bulged and supported by a transducer with an aperture size of 16.74 mm, and surface curvature is defined as an arc of 30-mm radius circle centered at (0, 31.3) mm, mimicking lateral focusing. Note that the curved transducer represents the linear transducer with a focus at the center of curvature, (0, 31.3) mm.

Table 1 Acoustical properties of simulated breast tissues and couplant.

Acoustical properties per tissue/material	Skin	Gel	Adipose
Speed of sound (m/s)	1600	1490	1450
Density (kg/m ³)	1060	1060	900
Dynamic viscosity (Pa s)	0.5	0	0.5
Bulk viscosity (Pa s)	0.5	0	0.5

Fig. 2 Simulated breast geometry consisting of couplant, skin, and adipose tissue.

2.2 Wave Propagation Theory

The wave propagation in lossy viscous media with respect to any given time t can be described by the following equation:²⁵

$$\frac{1}{\rho c^2} \frac{\partial^2 p}{\partial t^2} + \nabla \cdot \left[-\frac{1}{\rho} \nabla p + \frac{1}{\rho c^2} \left(\frac{4\mu}{3} + \mu_B \right) \frac{\partial \nabla p}{\partial t} \right] = 0, \quad (1)$$

where p is the total pressure that is dependent on the given speed of sound c , density ρ , dynamic viscosity μ , and bulk viscosity μ_B . The initial conditions for Eq. (1) are given as $p = p_0$ and $\partial p / \partial t = 0$ at $t = 0$, where p_0 is the initial pressure (see Table 2).

The transmitted pressure from the transducer can be modeled as the inward acceleration of the mesh element whose direction is normal to the transducer surface (red curve in Fig. 2) and points toward the simulation domain. This can be written as

Table 2 Constants used for simulating wave propagation.

Governing constant	Value
Initial pressure (p_0)	101325 Pa
Pulse delay (t_0)	3 μ s
Maximum inward acceleration (a_0)	10000 m/s ²

$$-\mathbf{n} \cdot \left(-\frac{1}{\rho} \nabla p \right) = a_n, \quad (2)$$

where p is the transmitted pressure, $\mathbf{n}$ is the normal vector of the transducer surface, and a_n is the inward acceleration. The inward acceleration pulse with respect to the lateral location x is given by a sinusoidal acceleration tapered by a Gaussian function and transducer apodization window

$$a_n(x) = a_0 H(x) e^{-[(t-t_0)1.25f_0]^2} \sin(2\pi f_0 t), \quad (3)$$

where f_0 is the center frequency and $H(x)$ is the Hamming window defining transducer apodization. The additional parameters a_0 and t_0 are given in Table 2.

In addition, to minimize reflections from the simulation domain boundaries, the following radiation boundary conditions are modeled:

$$-\mathbf{n}_b \cdot \left(-\frac{1}{\rho} \nabla p_b \right) + \frac{1}{\rho} \left(\frac{1}{c} \frac{\partial p_b}{\partial t} \right) = Q_i, \quad (4)$$

$$Q_i = \frac{1}{\rho} \left(\frac{1}{c} \frac{\partial p_i}{\partial t} \right) + \mathbf{n}_b \cdot \frac{1}{\rho} \nabla p_i, \quad (5)$$

$$p_i = e^{-\left(\frac{t-x/c}{0.01}\right)^2}, \quad (6)$$

where p_i is the incident pressure field, $\mathbf{n}_b$ is the normal vector of the boundary, Q_i is the acceleration of incidence field in the direction of $\mathbf{n}_b$ on the domain boundaries, x is the lateral location, and p_b is the pressure radiated out of the boundary to cancel reflections caused by simulation domain boundaries.

2.3 Simulation Parameters and Assumed Properties

The simulation studies were conducted using center frequencies of 1 and 4 MHz, each with three different choices of couplants. Such three couplants mimicked acoustical properties, i.e., speed of sound, density, and viscosity of commercial gel couplant,^{37,38} skin,³⁹ and adipose tissue.^{12,40,41} To achieve an attenuation coefficient of $\alpha = 0.6$ dB/cm/MHz, the dynamic and bulk viscosity of 0.5 Pa s was applied in skin and adipose studies.

The simulation domain was divided into meshes, where the mesh size was primarily determined by the medium's speed of sound, the used imaging frequency, and a factor that adjusts the mesh size in terms of a fraction of the local wavelength, called spatial sampling rate (SSR). Such parameters are given in Table 3. The maximum growth of neighboring meshes with respect to the given mesh is controlled by the maximum element growth rate. Furthermore, the maximum mesh size is controlled by the curvature factor: the mesh size must not exceed the curvature radius of the given geometric boundary, multiplied by the curvature factor. The parameters that constrain the change of mesh size and maximum mesh size allowed with respect to the geometric boundary curvature are given in Table 4.

The simulation was performed using the COMSOL multiphysics program V4.2 (COMSOL, Inc., Burlington, Massachusetts), which models wave propagation via the FDTD approach.

Table 3 Mesh properties used for 1 and 4 MHz simulation studies. Note that for computational time feasibility at higher frequency, the SSR per wave cycle was reduced to expand the maximum grid size limit.

Mesh property per center frequency f_0	$f_0 = 1$ MHz	$f_0 = 4$ MHz
Temporal sampling rate per wave cycle (TSR)	6	11
SSR per wave cycle	6	4
Maximum SSR per wave cycle (SSR _{max})	11	11
Maximum element size	$c/(f_0 \text{SSR})$	$c/(f_0 \text{SSR})$
Minimum element size	$c/(f_0 \text{SSR}_{\text{max}})$	$c/(f_0 \text{SSR}_{\text{max}})$

Table 4 Mesh properties used for different materials. To avoid aliasing due to spatial undersampling caused by higher speed of sound, note that the maximum element growth rate and the curvature factor in regions representing skin-like material were set to have slightly lower values than those in regions representing gel-like and adipose-like materials. Lowering these properties for gel-like and adipose-like materials, however, does not yield any accuracy improvement.

Mesh property per material	Skin-like	Gel-like	Adipose-like
Maximum element growth rate	1.1	1.3	1.3
Curvature factor	0.2	0.3	0.3

2.4 Simulations

2.4.1 Wavefront image generation

To visualize the beam patterns and progression across all simulation times (from 0 to 30 μs with the step sizes of 0.25 μs for 1 MHz studies and 0.0625 μs for 4 MHz studies), all wavefronts from all simulation times were superimposed on the same simulation domain. Specifically, the superimposed image, i.e., the wavefront image, was generated by the following: for any given grid point on the simulation domain, the pressure value was set to the one with the highest absolute value among all simulation times at the same grid point location, i.e., “max-hold” in time. Note that, from now on, we will define this superimposing procedure as max-hold method. Note that, before performing the max-hold operation, a median filter with a patch size of 0.75 mm $\times$ 0.75 mm was applied as some (<0.1% by area) of the simulated signals were artifacts with unusually high amplitude.

Twenty-one wavefront images were generated from 21 datasets corresponding to 21 different lateral beam positions. Specifically, the beam was moved laterally from 0 to 8 mm by 0.4 mm increments, thus simulating an image with a 400 μm lateral beam spacing.

2.4.2 Determining imaging beam deflections

The imaging beam deflection, i.e., the directional change of the wave from the original propagation direction due to an acoustic

Fig. 3 Beam deflection caused by acoustic impedance mismatch at the interface between two different media. The deflection angle (θ) can be obtained by calculating the difference between the directions of the initially propagated beam (solid line) and the deflected beam that has the highest spatial average positive peak pressure (dashed line). Figure labels and annotations are omitted for more clarity; see Fig. 2 for annotations.

impedance mismatch at the interface of two adjacent media, is determined by the wavefront by the most dominant beam refracted after traversing the interface. The most dominant beam is defined as the one with the highest spatial average positive peak pressure over the chosen positive wave peaks on the beam. The following paragraph explains how this average positive peak pressure is determined.

Since it is difficult to determine the most dominant beam *a priori*, several beams were initially considered. Then, for each such beam, the positive wave peaks were localized and the average pressure amplitude of the positive wave peaks was assessed. Only the peak locations from the beam with the highest average pressure amplitude were used to determine the beam propagation direction. Simple linear regression was employed to estimate the line-fitting lateral location of the particular peak given its axial location, and the calculated slope was used for calculating the deflection angle (see Fig. 3 for illustration of the deflection angle). The same procedure was repeated three times on the same beam, and the average of three deflection angles was calculated. For each type of couplant, the average

deflection angle was also calculated as the average of all deflection angle magnitudes among all transducer lateral shifts.

2.4.3 Determining signal dropout areas

To demonstrate the effects of signal dropout appearing over a wide range of imaging locations, 21 wavefront images reflecting shifts of 0 to 8 mm were first converted to positive values by the absolute operation ($|\cdot|$) and were then superimposed using the max-hold method. The field-of-view range of the final image containing the signal dropout area was set to -4 to 12 mm laterally and 5 to 40 mm axially, with the uniform resolution of 0.05 mm (see Fig. 2 for dimension details). Since there is a mismatch between the spatial distribution of pressure mesh point in wavefront images and the signal dropout area image, the pressure mesh point in the 2-D plane was interpolated onto a rectangular grid via the scattered interpolation method.

The resulting image is the max-hold superimposed pressure distribution from all lateral beam locations showing the signal dropout. The signal dropout area was enclosed by an isocontour whose amplitude was measured by 3 dB down from the minimum of two maximum signal pressures on the cross-section signal profile obtained at the top of the image (greatest axial depth). These two maximum points correspond to the maximum values on two signal profile segments separated by skin layers; see Fig. 4 for illustration. Once the isocontour of the signal dropout area was determined, the enclosed area was calculated; the minimum, maximum, and average values of all field points within the signal dropout area were also determined.

Note that the propagated signal was attenuated by $\alpha = 0.6$ dB/cm/MHz. For this reason, a time gain compensation (TGC) technique was applied to equalize the attenuated signal with depth. Knowing that the assuming propagation medium has a homogeneous speed of sound of c and a homogeneous linear attenuation of α , the correction factor V at the time t is given as

$$V(t) = e^{\frac{tcaf_0}{8.686}}. \quad (7)$$

Fig. 4 (a) Illustration of a signal profile corresponding to the horizontal cross section of the signal dropout area image at the greatest axial depth (40 mm). Two maximum signal pressures on the profile are marked as blue dots. The signal profile under the thick black dashed line represents part of the signal dropout area. All contour points (red) constitute the isocontour of the signal dropout area shown in (b) and Fig. 7.

Fig. 5 Wave propagation pattern as a result of different couplants, beam lateral positions, and two center frequencies. The resultant images are grouped by center frequency and couplant name. For each group, four images from left to right represent wave propagation patterns obtained at lateral beam positions of 0.4, 2.0, 4.0, and 7.6 mm, respectively.

3 Results and Discussion

3.1 Deflections of the Transmitted Beams

Figure 5 shows examples of beam refraction patterns that occur when the beam transverses the interface between couplant, skin, and breast tissue. Refraction is a function of the local impedance mismatch, the boundary geometry (function of the lateral beam position), and the center frequency. The refracted beam with the highest average pressure amplitude, i.e., the most dominant beam, is marked by a red trace. Figure 6 shows the corresponding deflection angle of the most dominant beam, with respect to a vertical path, at both center frequencies, the three couplants, and the 21 beam positions. Note that, due to unresolved technical issues, the simulation at 6.4 mm lateral beam position and 4 MHz center frequency could not be accomplished. The range of 0 to 8 mm was chosen because it is the range over which the artifacts due to the beam deflections are noticeable for the typical breast shape commonly found in clinical settings and chosen for the simulation. Also, the center frequencies of 1 and 4 MHz were chosen because there is a major transition of beam propagation behavior: between these two frequencies, increasing the center frequency from 1 to 4 MHz narrows the beam, which is then able to funnel within the skin layer. Positive deflection angles denote the wave refraction toward

the positive lateral direction (i.e., to the right side, see Fig. 3 for illustration), and negative deflection angles denote the wave refraction toward the negative lateral direction. In general, the deflection curves generated from 1 and 4 MHz datasets [Figs. 6(a) and 6(b), respectively] have somewhat similar characteristics; in the beam shift range of 0 to 1.6 mm, the beams deflect into the breast adipose tissue, and in the range of 4.4 to 8 mm, the beams deflect away from the breast. However, the center frequency seems to play a role in how beams are deflected outside the mentioned ranges. This effect can be seen from both Figs. 5 and 6(b) when commercial-gel-like or adipose-like couplant is used and the center frequency is set to 4 MHz. The beam appears to funnel within the skin layer. This effect, however, does not appear with the lower, 1 MHz, center frequency. This could be explained by the fact that the beam-focusing capability is enhanced when a higher frequency is used, i.e., a narrower beam width is realized and therefore most of its energy can be confined and funneled within the skin layer.

Also, as expected, using the skin-like couplant results in the least amount of beam deflection. However, if the most dominant beam lies inside the breast tissue, the skin-like couplant will cause the greatest negative deflection compared with commercial-gel-like and adipose-like couplants for both 1 and 4 MHz center frequencies, as can be seen from Fig. 6. This could be explained as the beam traveling through the skin-like couplant

Fig. 6 Deflection angle of the most dominant beam as a result of different couplants and lateral beam positions. (a) and (b) correspond to the used center frequencies of 1 and 4 MHz, respectively. Two important aspects to note at a particular lateral transducer shift are (1) for beam position not >4 mm, the skin-like couplant will cause the greatest negative deflection, slightly greater than commercial-gel-like and adipose-like couplants, the direction causing signal dropout in the breast. There is less positive deflection for beam position >4 mm with the skin-like coupling, but that angle is basically irrelevant and (2) using the higher center frequency results in more complex beam deflection patterns, especially with the commercial-gel-like and adipose-like couplants.

Fig. 7 Signal dropout area as a result of using three different couplants, simulated at center frequencies and shown at dynamic ranges as follows: (a) 1 MHz and 40 dB range, (b) 4 MHz and 40 dB range, (c) 1 MHz and 72 dB range, and (d) 4 MHz and 72 dB range. The color bar represents the pressure amplitude in dB, where the maximum signal is set to 0 dB. The blue isocontour can be determined by the approach demonstrated in Fig. 4. Legends S, G, and A represent skin-like, commercial-gel-like, and adipose-like couplants, respectively.

will deflect only at the interface between the inner skin layer and the adipose tissue in the breast, and, since the adipose tissue has a lower speed of sound than the skin, this refraction will cause the beam to deflect to the left side (negative angle). On the other hand, using either the commercial-gel-like or the adipose-like couplant that has lower speed of sound than skin will initially cause the beam to deflect to the right side when the beam is hitting the outer skin layer. Note that this deflection, in turn, causes a greater incidence angle of the beam at the interface between the inner skin layer and the adipose tissue than the same incidence angle resulted from using the skin-like couplant, thus making the final beam bend less toward the left side. On the other hand, when the adipose-like couplant is used, it is important to note that the beam deflections in the breast seem to be greater toward the positive direction than those resulted from using commercial-gel-like couplant. The lower speed of sound of adipose tissue causes the greater positive deflection angle of the beam at the outer skin layer. The greater the positive deflection angle at the outer skin layer, the greater the positive incidence angle at the interface between the inner skin layer and adipose tissue, and this greater incidence angle will cause the final beam to bend more toward the right.

3.2 Areas of Signal Dropouts

Figure 7 shows the results of signal dropout area in the simulated breast using various couplants. The signal dropout area

characteristics (size, minimum, maximum, and average of all pressure samples in the signal dropout area) are given in Table 5 and the box plots in Fig. 8. As can be seen from both Figs. 7–8 and Table 5, both the center frequency and the couplant affect the size and the signal intensity in the dropout area.

When using the higher imaging frequency (4 MHz), the propagated beam will be narrower and allow for better spatial resolution, but higher attenuation due to higher frequency can cause a weaker average signal intensity in the dropout area, as suggested by Table 5 and Fig. 8. Straightforward solutions to compensate for the signal loss are to increase TGC or overall gain (see Sec. 2.4.3 for more details) or to increase the dynamic range. Unfortunately, neither solution is capable of mitigating the signal dropout area. The TGC technique is only able to recover the signal loss due to the attenuation but not the refraction due to the impedance mismatch at the interface between two media. Increasing the dynamic range helps visualize the image regions with low intensity as demonstrated in Fig. 7, but it is impractical in situations of low SNR.

It is observed that, to decrease the size of the signal dropout area, it may be better to choose an appropriate couplant instead of pursuing the aforementioned solutions. Couplants affect the signal dropout area characteristics due primarily to phase interference resulting from the speed of sound and acoustic impedance mismatches. As can be seen, both Fig. 7 and Table 5 suggest that the skin-like couplant results in the smallest signal

Table 5 The properties of signal dropout area as results of using different couplants and center frequencies.

Center frequency	1 MHz			4 MHz		
Coupling material	Skin-like	Gel-like	Adipose-like	Skin-like	Gel-like	Adipose-like
Average absolute deflection angle (deg)	9.0	14.6	20.6	12.0	15.1	20.3
Dropout area (mm ²)	21.0	49.0	62.0	1.00	30.0	41.0
Minimum signal intensity in dropout area (dB)	−11.1	−20.5	−23.7	−20.0	−28.4	−25.7
Maximum signal intensity in dropout area (dB)	−6.54	−8.54	−8.33	−18.4	−20.6	−18.6
Average signal intensity in dropout area (dB)	−8.03	−12.0	−12.7	−19.1	−24.7	−22.5

Fig. 8 Box plots of signal intensity in dropout area, as results of using (a) 1 MHz and (b) 4 MHz center frequencies.

dropout area, followed, in order, by the increasingly lower speed of sound commercial ultrasound gel and adipose-like couplants. In addition, it is interesting to note that the area of the signal dropout seems to be related to the average absolute deflection angle obtained across all beam lateral shifts. As can be seen from Table 5, the average absolute deflection angle with the three couplants follows the same order as the area of the signal dropout. The reason for using skin-like couplant resulting in the smallest signal dropout area is obviously not only the reduction of one effective interface but also the reduction of absolute speed of sound difference with the skin, which causes a more advantageous refraction result. Note that we have not attempted to make an empirical or analytical fit to these results, as details of the effects will vary in a complex way depending on the particular geometry and acoustic properties of the tissues.

Furthermore, as can be seen from Table 5 and Fig. 8, when a skin-like couplant is used, the average signal intensity in the dropout area is greater and the range of signal intensity in the dropout area is narrower than those obtained when using a gel-like or adipose-like couplant. This could be attributed to the reduction in acoustic impedance mismatch that helps ablate overall beam deflection as a result of using the skin-like couplant. Thus, it suggests that using the couplant that minimizes the acoustic impedance mismatch not only improves the signal intensity within the dropout area but also helps eliminate the circumstances where low and high intensity areas appear closely to each other and might confuse radiologic interpretations. The advantages of replacing the gel with a skin-like couplant will be more noticeable in pulse-echo imaging, where round-trip wave propagation occurs as opposed to the one-way wave propagation simulated in this paper.

It is also expected that, for frequencies lower than 1 MHz, the signal intensity within the dropout area will increase due to reduced attenuation; thus, the artifacts will diminish. For frequencies higher than 4 MHz, the dropout areas should appear similar to those of the 4 MHz case as beam deflection patterns will be similar for 4 MHz and above.

The use of relatively low frequencies in the range of transmission tomography systems, compared with the usual 8 to 15 MHz of manual breast imaging, was necessary for computational efficiency. However, the results at frequencies differing by a factor of three give some suggestions of trends at higher frequencies.

4 Conclusions

This paper has demonstrated how acoustic impedance mismatch between coupling material, human skin, and subcutaneous breast tissues leads to some of the anomalies seen in imaging, such as during automated whole breast ultrasound and transmission tomography, where the transducers are not always directly in contact with the skin. These anomalies manifest themselves as signal dropout areas, or shadows, inside the breast when the imaging beam transverses the skin at a fairly steep angle (40 deg to 60 deg). This would be particularly important in the subareolar region, where cancer is relatively common near the skin.^{35,36} The simulations showed that reducing the acoustic impedance mismatch between skin and couplant decreases the area of and signal intensity loss in the signal dropout area. Even though only 1 and 4 MHz frequencies were used in the simulations, we expect that this main conclusion will still hold for diagnostic frequencies, as we have seen similar types of artifacts in clinical images, as shown in Fig. 1. Despite that subcutaneous fat was always assumed beneath the skin, in rare other

cases, the subareolar region located close to the skin might also have a speed of sound close to that of the skin. Therefore, similar, possibly increased, artifacts could also be expected around that region, and using couplant with skin-like acoustic properties could then mitigate such artifacts.

Design of an ultrasound couplant with acoustic properties close to those of breast skin might not be trivial. One of these possibilities might be to utilize an ethanol–water mixture with a 17% concentration by weight, which has been shown to provide a speed of sound close to that of breast skin (1605 to 1610 m/s) and provide a negligible temperature coefficient of speed of sound in the range of room to body temperatures.⁴² However, control of the volatility of the ethanol might be needed. Modest reduction in the temperature of the breast and couplant would also reduce the speed of sound contrast in the breast and resulting artifacts, but there are clear limitations related to patient comfort.

It was observed that using the skin-like couplant could cause slightly greater beam deflection within the breast, but this increase in deflection is outweighed by the considerable decrease in the signal dropout area compared with using commercial ultrasound or adipose-like gel.

When performing compound imaging in a breast freely suspended in a coupling medium, without having transmission data and doing three-dimensional (3-D) full-wave migration, this type of analysis would potentially allow the exclusion of erroneous data from compound summations. One might also include some refracted wave data after approximate corrections for the expected bending of wavefronts. The surface of the skin should be tracked in 3-D in the images performed up to the image being formed to make these corrections in, eventually, real-time interpolations of a more extensive set of data than that provided here. This correction should result in much improved image quality at distances of up to several centimeters from the skin.

Short of full-wave migration imaging, wave migration calculations in a limited volume might be performed to allow corrections around recognizable and segmentable internal structures for which reasonable estimates of the speed of sound could be made.^{43–45} This could strongly reduce shadows at the edges of structures with smooth borders, such as suspensory ligaments, cysts, and fibroadenomas, revealing possible diagnostic information about poorly seen distal borders and providing more information about the attenuation properties of the tissues of the smooth-walled masses themselves, independent of their speed of sound differences with the surrounding material.^{16,17} Shadow reduction might be less with the majority of invasive carcinomas, those having very diffuse borders.

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