Published in final edited form as:

Ultrasound Med Biol. 2014 July ; 40(7): 1697–1717. doi:10.1016/j.ultrasmedbio.2014.02.011.

LOW-ECHO SPHERE PHANTOMS AND METHODS FOR ASSESSING IMAGING PERFORMANCE OF MEDICAL ULTRASOUND SCANNERS

Ernest L. Madsen, Chihwa Song, and Gary R. Frank

Department of Medical Physics, University of Wisconsin, Madison, Wisconsin, USA

Abstract

Tissue-mimicking phantoms and software for quantifying the ability of human observers to detect small low-echo spheres as a function of depth have been developed. Detectability is related to the imager's ability to delineate the boundary of a 3-D object such as a spiculated tumor. The phantoms accommodate a broad range of transducer shapes and sizes. Three phantoms are described: one with 2-mm-diameter spheres (for higher frequencies), one with 3.2-mm-diameter spheres (for lower frequencies) and one with 4-mm-diameter spheres (for lower frequencies). The spheres are randomly distributed in each phantom. The attenuation coefficients of spheres and surroundings are nearly identical; thus, compromising shadowing or enhancement distal to spheres does not occur. Reproducibility results are given for pairs of independent data sets involving eight different combinations of scanner, transducer and console settings. The following comparison results are also reported: (i) only the selected frequency differs; (ii) transducers and scan parameters are nearly the same but manufacturers differ; (iii) ordinary B-scanning, spatial compounding and tissue harmonic imaging are addressed. The phantoms and software promise to be valuable tools for scanning system and setup comparisons and for acceptance testing.

Keywords

Phantom; Imaging performance; Software; Low-echo spheres; Detection

INTRODUCTION

One way to assess the ability of an ultrasound scanner to delineate the boundary of a 3-D object such as a spiculated tumor is to determine its ability to allow a human observer to detect the presence of a small low-echo sphere throughout the image. Phantoms containing low-echo spheres with co-planar centers forming a regular array have been reported (Kofler and Madsen 2001; Kofler et al. 2005). The spheres are low-echo with respect to the surrounding background; software for quantifying the human observer's ability to detect these spheres was also reported. That software was an adaptation (Kofler and Madsen 2001;

Kofler et al. 2005) of a method applying to a cylinder with its axis perpendicular to the scan plane (plane of symmetry of the scan slice) reported previously (Lopez et al. 1992).

A version of the phantom with the regular array of co-planar spheres is available commercially (Catalog No. 408 LE, Gammex, Middleton, WI, USA). That phantom has a flat scanning window limiting its applicability for convex arrays. Production of such a phantom applicable for a convex array with little restriction on the radius of curvature (ROC) would likely be too costly.

We report on phantoms with conical and flat scanning windows which allow low-echo sphere detectability to be quantified for any transducer geometry, including convex arrays with ROCs from 0.5 through 6.5 cm. Each phantom contains identical low-echo spheres—all with the same diameter—randomly distributed spatially throughout the volume of the phantom.

User-friendly software quantifies detectability as a function of depth in three steps: (i) The centers of detectable spheres are determined using a set of image frames with parallel scan planes at elevational increments of one-fourth of the sphere diameter. (ii) The detectability of each sphere is expressed as the lesion signal-to-noise ratio (LSNR), which is determined using a variation of the method reported previously (Kofler and Madsen 2001; Kofler et al. 2005). (iii) The mean LSNR over depth increments is plotted versus depth.

Three sphere diameters have been investigated: 2, 3.2 and 4 mm. The 2-mm-sphere phantom is for use with higher nominal frequencies (7–15 MHz), and the 3.2- and 4-mm-sphere phantoms are for use at lower frequencies (2–7 MHz).

To provide for assessment over the entire image frame of convex and phased arrays while maintaining a practical size of the phantoms, parallel planar reflectors producing total internal reflection of incident compressional waves (Lamkanfi et al. 2009) are part of each phantom. Experimental tests for angles of incidence providing total internal reflection using an *ad hoc* phantom are reported in Appendix I.

These phantoms and associated software can be used for comparing different scanner and transducer combinations and related configurations for each. (The term *configuration* refers to all parameters selected by the user, such as nominal frequency, dynamic range, depth of field, time gain compensation and overall gain.) Another application is use as part of an acceptance-testing procedure.

Significance of the mean LSNR value to detectability of spheres by a human observer

There is a one-to-one correspondence between human observer detection performance and the mean LSNR at mean LSNRs ≥ 5 . The human observer detection performance is given in terms of fraction correct in a two-alternative-forced-choice (TAFC) experiment, as described previously (Kofler et al. 2005). A score of 1 corresponds to no error in detection; that is, the human observer always chose the correct of two images, one of which contains the ultrasound image of the sphere. A score of 0.5 means that the observer was completely

unable to detect the sphere, and the corresponding mean LSNR value is 0. The more negative the mean LSNR value, the greater is the detectability.

Figure 1 illustrates curve-fit results from Figure 7b of Kofler et al. (2005). The curves for 2-, 3- and 4-mm spheres are a little different, which may be due to sample size limitations. Note that for LSNR values less than (more negative than) -5 , the TAFC values are nearly 1 for all three sphere sizes. Looking at the graphs of mean LSNR versus depth in the Results section, generally much of each curve corresponds to values less than -5 . It is reasonable to assume that at each depth interval, the better (more negative) the mean LSNR is, the better the mean LSNR value would be for spheres with higher backscatter coefficients relative to that of the low-echo spheres in the phantom. Given a mean LSNR value -5 of low-echo spheres in a given depth interval, it may be possible to predict the backscatter coefficient of sphere material that would give rise to a mean LSNR of, for example, -2 under the same conditions. An experiment is proposed at the end of the Discussion and Summary that might give rise to a means for such a prediction.

METHODS

Phantom description

Phantoms containing 3.2- or 4-mm diameter spheres have almost the same geometry. Figure 2 is a photograph and Figure 3 is a diagram of the 3.2-mm-sphere phantom. The depth of the tissue-mimicking (TM) material perpendicular to the plane of Figure 3 is 16 cm. A photograph and diagram of the 2-mm-sphere phantom are provided in Figures 4 and 5, respectively. The depth is 10 cm for the 2-mm-diameter sphere phantom, and the depth is 20 cm for the 4-mm-diameter sphere phantom.

Sphere number densities—For each phantom, the number of spheres per unit volume has been chosen to be small enough that adequate background TM material is present for implementation of the data reduction method (see Data Reduction), but high enough that data acquisition is facilitated. The volume percentage occupied by spheres is 3.3% for each phantom. Thus, number of spheres per unit volume in the 4-, 3.2- and 2-mm-sphere phantoms is 1/mL, 2/mL and 8/mL, respectively. If the volume of TM material in each phantom is used, the total number of spheres is 4430 in the 4-mm-sphere phantom, 6720 in the 3.2-mm-sphere phantom and 10,400 in the 2-mm sphere phantom.

Tissue-mimicking materials and their introduction into the phantom container

Tissue-mimicking materials are versions of those reported (Madsen et al. 1998). The materials in the spheres and surroundings have the same composition except that the background contains 6 g/mL 45- to 53- μ m-diameter glass bead scatterers. Other than the glass beads, the materials are composed of reverse osmosis water, agarose (Agarose I, AMRESCO, Solon, OH, USA), milk concentrated by a factor of 3 in an ultrafiltration unit (Model UFP-10-C-55, A/G Technology, Needham, MA, USA), propylene glycol and Liquid Germall Plus, a preservative (ISP Technologies, Wayne, NJ, USA).

The procedure for producing approximately 1 L of TM material in liquid (molten) form and then congealing it to solid form is as follows. First, produce about 700 mL of agarose

solution. In a 1.5-L Pyrex beaker, mix 651 mL of reverse osmosis water with 49 g of propylene glycol. While stirring, add 28 g of agarose (dry weight). Cover the beaker with a piece of plastic wrap and secure with a rubber band. Poke a small hole in the plastic wrap to allow pressure to remain at atmospheric during heating. Place the beaker in a larger container (such as a bucket) of water so that the agarose mixture is below the external water level. The bottom of the beaker should rest on something to keep it from contacting the bottom of the container. Place on a heating coil and, while stirring occasionally, raise the temperature to at least 90°C, at which point the mixture becomes a transparent solution. The purpose of the container of water is to avoid heating the bottom of the glass beaker to a temperature much higher than the boiling temperature of water; such higher temperatures could “burn” the agarose in contact with the beaker. Cool the agarose solution to about 60°C, and mix 585 mL with 415 mL of 60°C 3x concentrated milk that was previously concentrated via ultra-filtering. (Note: The 3x concentrated milk should have been previously filtered at room temperature through a filter with a 10- μ m sieve-opening filter.) Cool to 50°C and add 8.78 g of Liquid Germall Plus, also at 50°C. This mixture is ready for production of low-echo spheres. The mixture will congeal at about 38°C. If background material is to be produced, then 6 g of 45- to 53- μ m-diameter glass beads at 50°C should be added to 1 L of 50°C mixture before congealing.

To produce the large number of spheres needed, two-part acrylic molds with opposing hemispherical depressions are used in our lab, the two parts being lowered into the 50°C molten sphere TM material and brought together to form spheres. (Constraining brass pegs and holes in the molds ensure sphere formation.) The two parts are clamped together and placed in a cold water bath for rapid congealing. Figure 6 is a photograph of a two-part mold for producing 1044 spheres.

The next step is to fill the phantom container with molten TM background having randomly distributed solid spheres in it. In one corner of the phantom container is a cylindrical hole with a sawed-off plastic syringe barrel epoxied in it. External constraints are placed over the flexible windows to maintain their shape. The spheres needed are warmed to 50°C and added to a volume of 50°C molten background material that is a few hundred milliliters less than that need to fill the container; this material is then poured into the phantom container through the syringe barrel. Additional molten background TM material is added and the syringe piston is inserted into the piston barrel so that no air bubbles exist in the molten TM material; rubber bands produce a force on the piston so that a positive gauge pressure exists in the TM material during congealing. The phantom is then oscillated by hand in a rotational fashion for 10 or 20 s to ensure random distribution of the spheres, and the entire phantom is clamped onto a device for rotating it uniformly about 2 revolutions per min about a horizontal axis. That rotation ensures that the 45- to 53- μ m glass beads remain uniformly distributed in the background material during congealing.

The TM material corresponding to the spheres has a density of 1.038 ± 0.005 g/mL, and that of the background is greater by only 0.1%; thus, the difference is slight. (Note that the materials constituting the background and spheres are identical except that the background contains 6 g of glass bead scatterers per liter of sphere material.) Sphere-to-background material relative backscatter was determined to be -30 dB by comparison of images of

cylindrical test samples of sphere and background materials. Measurements of propagation speeds and attenuation coefficients are summarized in Table 1; the methods of measurement have been previously reported (Madsen et al. 1999).

Data acquisition

Once the scanner, transducer and console settings have been selected for performance assessment, a set of images is obtained in a manner that facilitates determination of the positions of detectable spheres in the phantom. The gain and time gain compensation settings should be set so that the mean gray level of the background is reasonably uniform and is representative of clinical images. The method used is to translate the transducer incrementally in the elevational direction with step size $D/4$, where D is the diameter of the spheres in the phantom.

Photographs of the device for holding the transducer in a fixed position and then translating it semi-automatically in the elevational direction while maintaining contact with the scanning window are provided in Figure 7a and b. A manual translator (Part No. A1506K2-S1.5, Velmex, Bloomfield, NY, USA) provides an 11.4-cm vertical translation range of the transducer, and a stepper motor (Model No. 17PM-K858-00VS, 200 steps/mm, Minebea Motor, Tokyo, Japan)-driven translator (Part No. MA1504.5K1-S1.5, Velmex) provides a 7.6-cm range of horizontal translation of the transducer on the scanning window. A convex transducer with a ROC of about 5 cm is clamped between two metal plates covered with somewhat pliable silicone pads. The transducer is in contact with the section of the cylindrical scanning window that fits to the transducer emitting surface firmly. Also, there is considerable coupling gel in place. The direction of translation is parallel to the aluminum plate on which the phantom sits and toward increasing radius of curvature of the scanning window. Coupling gel and flexibility of the scanning window provide close acoustic coupling between the emitting surface and the scanning window during translation. The stepper motor (visible in Fig. 7b) is controlled by electronics in the black box on the left of the phantom in Figure 7a. When a button on the box is pressed, the transducer is moved the $D/4$ increment, after which an ultrasound image is recorded. The button is pressed again and another image recorded, and so on. The increment choices are 0.5, 0.8 and 1.0 mm, selected by pressing one of the buttons on the box an appropriate number of times.

The number of image frames needed depends on transducer geometry, frequency and focus depth. That number is addressed under Data Reduction.

Data reduction

Data reduction involves two distinct parts: determination of sphere centers and computation of the mean LSNR as a function of depth.

Determination of sphere centers—Consider each image to be subdivided into contiguous areas of depth d . For linear arrays, these areas will be rectangles, and for convex arrays, they will be sections of annular rings, the angular limits being determined by the sector angle. For phantoms with 3.2- or 4-mm-diameter spheres, depth intervals of 5 mm are

appropriate, and for phantoms with 2-mm-diameter spheres, depth intervals of 2 mm are appropriate.

The determination of sphere centers can be accomplished via the following steps.

1. Define MPV to be the mean pixel value over an area A in a phantom image, where A is a square with side $(2/3)D$, approximating the projected area of a sphere (Kofler et al. 2005). For each image recorded, a set of MPVs is computed with centers comprising a simple square array over the image area with nearest-neighbor spacing of $D/4$. Note that because the spacing is shorter than the side of the square ($D/4 < 2D/3$), the square areas A overlap as depicted in Figure 8. Thus, for all images obtained (separated by $D/4$), each MPV is associated with one of the sites in a simple cubic array.

Define $(MPV)_{ijk}$ to be the MPV at the ijk site, where k corresponds to the elevational (translational) direction and to the z -axis of a Cartesian coordinate system, and i and j correspond to vertical and horizontal axes on an image.

2. For each depth interval d , compute the mean M_d and standard deviation SD_d of MPVs using the entire image set.
3. For each MPV_{ijk} , identify six nearest MPVs and determine whether all seven MPVs, including itself, are at least 1.5 standard deviations below the mean established in step 2. Each such set of seven is taken to be associated with a low-echo sphere. (See Appendix II for the reason for choosing the seven specified sites.)
4. Typically, a set of seven MPV values determined in step 3 will have common positions with other such sets of seven found in step 3. Associate all such sets with one sphere.
5. For sets of seven MPVs corresponding to step 4, compute the best estimate for the position of the sphere center by computing the Cartesian coordinates of the “center of mass” (x_{CM} , y_{CM} and z_{CM}) as in equations (1)

$$\begin{aligned} x_{CM} &= \frac{\sum_{ijk} (i) (D/4) [M_d - (MPV)_{ijk}]}{\sum_{ijk} [M_d - (MPV)_{ijk}]} \\ y_{CM} &= \frac{\sum_{ijk} (j) (D/4) [M_d - (MPV)_{ijk}]}{\sum_{ijk} [M_d - (MPV)_{ijk}]} \\ z_{CM} &= \frac{\sum_{ijk} (k) (D/4) [M_d - (MPV)_{ijk}]}{\sum_{ijk} [M_d - (MPV)_{ijk}]} \end{aligned} \quad (1)$$

Automated data analysis for quantifying sphere detectability—The correspondence of LSNR values to human observer detectability via TAFC methodology has been reported (Kofler et al. 2005). Thus, the human observer-related detectability equals the mean LSNR in each one of contiguous depth intervals spanning the entire depth range available. The LSNR is a numerical value quantifying the detectability of a macroscopically uniform target object in a macroscopically uniform surrounding background. Object and background materials have intrinsic backscatter coefficients, that is, BSC_{obj} and BSC_{bkg} , respectively. (Note that the “object contrast: in decibels equals $10 \log[BSC_{obj}/BSC_{bkg}]$, and it is recommended that this value be -20 dB or lower for the sphere and background materials.)

Elimination of all MPV values that might be influenced by the presence of a sphere—For computation of background means and standard deviations, it is important to not use any cubic array sites that lie within $(3/4)D$ of any sphere center.

Computation of the mean LSNR value for a given depth interval—The LSNR value for the n th sphere with center at $(x_{CMn}, y_{CMn}, z_{CMn})$ in a depth interval d is defined by

$$LSNR_n = \frac{(S_{Ln} - S_{mBn})}{\sigma_{Bn}} \quad (2)$$

where S_{Ln} is the MPV value with x and y coordinates equal to x_{CMn} and y_{CMn} in the image having z coordinate closest to z_{CMn} ; S_{mBn} is the mean of all MPVs in that image that are within a radius $2D$ of x_{CMn}, y_{CMn} and also are not influenced by the presence of any sphere including that centered at $x_{CMn}, y_{CMn}, z_{CMn}$ (see previous subsection); σ_{Bn} is the standard deviation of all MPVs contributing to S_{mBn} .

The mean LSNR value for the depth interval is then

$$\text{mean LSNR} = (1/N) \sum_{n=1}^N LSNR_n \quad (3)$$

where N is the total number of spheres detected in the depth interval (including the entire image set).

Standard errors—The standard error corresponding to each mean LSNR value is given by $N^{-1/2}$ times the standard deviation of the $LSNR_n$ values corresponding to the depth interval involved.

Curve fit method—Mean LSNR values versus depth are fitted by a rational function that is the ratio of polynomials given by

$$f(x) = \frac{p_1 x^2 + p_2 x + p_3}{x^2 + q_1 x + q_2} \quad (4)$$

where p_1, p_2, p_3, q_1 and q_2 are free parameters.

Estimate of the minimum number of images needed for acceptable results

Experience indicates that the minimum number N of spheres detected per depth interval in the focal zone should be about 25 to ensure acceptable results. The number of images in the data set, N_I , needed to obtain an average N of 25 is

$$N_I = \frac{25}{S \times d \times (L - 4D)D/4} + 2 \quad (5)$$

where $S \equiv$ the average number of spheres per milliliter, $d \equiv$ depth increment, $L \equiv$ lateral extent of the depth interval and $D \equiv$ sphere diameter. For linear arrays, $L =$ the width of the depth interval (usually the lateral width of an image), and for convex arrays, $L = 2\pi(F + R) \times (\alpha/360^\circ)$, where F is the focus depth (smallest focus depth when there is more than one), and R is the radius of curvature of the convex array. The $-4D$ term in the denominator of the right side of eqn (5) is present because the software eliminates all detected spheres having centers within $2D$ of the ends of L so that background MPVs will be adequately represented in the computation of S_{mBn} and σ_{Bn} . The “+ 2” term on the right side is present because the method for locating sphere centers precludes the location of a sphere center in the first and last images in an image data set.

As an example, consider a convex array with a radius of curvature of 5 cm. If the number of spheres per milliliter in the phantom is 2, the depth interval is 5 mm (0.5 cm), the focus is at 4 cm, the sector angle is 70° and D is 3.2 mm (0.32 cm), then $N_I \approx 34$.

RESULTS

Reproducibility and adequacy of 25 spheres per depth interval

A test of the phantoms and the software is whether two independent image sets produce the same result. Independent image sets are those that involve different volumes of the phantom; for example, if the number of consecutive images obtained is 70, then images 1–35 and images 36–70 involve completely different volumes of the phantom. If 25 is an adequate value for N_I , the results should agree.

4-mm sphere phantom

Two examples are presented, one for a convex array and one for a linear array. The first example involves the convex array. Figure 9 is one image from an image set. The nominal frequency was 4.15 MHz, and the focus is at 4 cm. Figure 10 relates to some aspects of the software application. Fifty images with elevational separation of $D/4 = 1$ mm were acquired for this example, and results are illustrated in Figures 11 and 12.

The second example involves a linear array operated at 4 MHz and focused at 3 cm. Figure 13 is an image, Figure 14 illustrates determination of sphere centers and Figures 15 and 16 address reproducibility.

3.2-mm phantom with one case in which only focusing is changed—The first example is for a linear array operating at 4 MHz with a focus at 3 cm. The elevational interval between images is $3.2 \text{ mm}/4 = 0.8 \text{ mm}$. An image is provided in Figure 17. Results for two cases in which there is no overlap between imaged volumes are illustrated in Figure 18. Both sets contain 70 images corresponding to a net elevational displacement of $69 \times 0.8 \text{ mm} = 5.52 \text{ cm}$. Here, the mean number of sphere centers per 5-mm depth interval is about 15 instead of 25, but reproducibility is still good. Figure 19 illustrates the result when the entire 140 images are used, and the average number of sphere centers per depth interval is about 30; the curve fit is excellent, indicating excellent reproducibility.

The next reproducibility study involves a convex array operating at 4.5 MHz and illustrates the ability of the system to differentiate performance for different focusing choices. Figure 20 is an image with multiple lateral foci at 4, 8 and 12 cm, and Figure 21 is a graph of mean LSNR values versus depth for this multiple-focus case. Figure 22 corresponds to a single deep lateral focus at 10 cm, and Figure 23 corresponds to a single shallow focus at 4 cm. The study indicates the dominant effect of fixed elevational focusing at a depth of about 4 cm.

2-mm sphere phantom

Figure 24 is an image of the 2-mm-sphere phantom made with a 1.5-cm-ROC convex array, and Figure 25 illustrates reproducibility results with lateral focus at 3 cm. This study again indicates a dominance of elevational focusing. Figure 26 illustrates the improvement when the entire 100-image set is employed, resulting in more than 25 spheres detected per depth interval in the apparent focal depth range. It was not reasonable to obtain more than about 100 consecutive images because the ROC of the conical window was becoming too large for good contact between transducer and window to be maintained.

Figure 27 is the image of the 2-mm-sphere phantom made with a high-frequency linear array with lateral focus at 4 cm. The results are illustrated in Figures 28 and 29. The dominance of elevational focusing is once again illustrated. Reproducibility was excellent.

Comparison studies

Change in transducer frequency only—Figures 30–32 illustrate results for the Siemens S2000 with an 18L6 transducer operating in standard B-mode with a focus at 4 cm, that is, no compound imaging, tissue harmonic imaging, persistence, and so on; however, the nominal frequencies are 8, 10 and 15 MHz. Surprisingly, very little difference is observed between the three cases.

Standard B-mode imaging and two processing techniques—This study was repeated using the Siemens S2000 with the 18L6 transducer. The nominal frequency was 15 MHz. The result for standard B-mode imaging is illustrated in Figure 32, and the results using spatial compound imaging (Siemens SieClear 5) and tissue harmonic imaging are illustrated in Figures 33 and 34, respectively. Considerable improvement in performance is apparent in terms of the mean LSNR versus depth for both spatial compounding and tissue harmonic imaging.

Two different makes of scanner with similar transducers and console settings

—Standard B-mode images were assessed for a focus at 4 cm and convex arrays with nearly the same sector angle. Also, the console settings were the same except for a small difference in nominal frequency. Figure 35 illustrates results for the Siemens S2000 and 7CF2 3-D transducer operated at 4.5 MHz in 2-D mode, and Figure 36 illustrates results for the Ultrasonix “Sonix Touch” scanner with a 4DC7-3 3-D transducer operated at 4 MHz in 2-D mode. The Siemens system appears to considerably outperform the UltraSonix system.

DISCUSSION AND SUMMARY

These phantoms and software allow quantitative analysis of the imaging performance for any shape of transducer emitting surface with respect to the ability to delineate the boundary of a 3-D object. The examples presented indicate that with sufficient data, definitive comparisons can be made between different scanner-transducer combinations and console settings. Use for acceptance testing should also be of considerable value.

Two phantoms were employed for use at lower frequencies. Because the volume of a 3.2-mm sphere is four times that of a 2-mm sphere, and the volume of a 4-mm sphere is eight times that of a 2-mm sphere, the 3.2 mm sphere phantom seems adequate for lower frequencies.

The amount of time need to acquire data is about 7 min. Our experience is that once a “new patient” folder has been created and the desired console settings have been selected, it takes perhaps 4 min to clamp the transducer in the translation apparatus and position it with coupling gel on the scanning window. Note that it is not necessary to take great care trying to make the scan plane exactly perpendicular to the translation direction; an angle of 10° or less between the translation direction and the perpendicular to the scan plane is sufficient ($\cos 10^\circ \approx 0.98$). Recording a 60-image data set requires about another minute. The image set can then be downloaded onto a flash drive in about 2 min for transfer to a computer for analysis.

Regarding data analysis, about 6 min is required to generate the graph of mean LSNR versus depth. The analysis programs (one for linear arrays and one for sector scanners) exist in executable graphical user interface form for use with a 64-bit computer running Windows 7 or 8. Setting up for data input takes about 2 min. Data input takes about another minute, detecting spheres another minute and computing mean LSNR values perhaps another 2 min.

The materials used to produce the phantoms are those used to produce most ultrasound phantoms by Gammex. According to the production engineer at Gammex, the expected shelf life of one of these phantoms is about 10 y, assuming that the user returns the phantom for simple replacement of solution when the weight decreases by a specified percentage.

An interesting observation from the limited number of scanner-transducer combinations and console settings is that the elevational focusing appears to dominate detectability more than expected when the elevational focus depth is fixed. A future article is planned to address transducers providing some elevational focusing *e.g.*, “1.5-D” arrays and, of course, 2-D arrays.

Possible estimation of mean LSNR values for sphere material with higher backscatter coefficients

Future work will include experiments with a set of phantoms having the same background backscatter coefficient, but various backscatter coefficients of the sphere material; that is, one phantom will have spheres with no added scatterers and the other phantoms will have spheres with different concentrations of glass bead scatterers. The diameter distribution of the glass bead scatterers will be the same for background and spheres. Mean LSNR versus depth will be generated for a variety of scanner/transducer configurations. The sphere material backscatter coefficients relative to that of the background (object contrast) might be -30 , -24 , -18 , -12 and -6 dB for a total of five phantoms. A result of these experiments might be that given the mean LSNR at some depth using one of the low-echo-sphere phantoms described in this report, the mean LSNR values for lower-object-contrast (absolute value) sphere phantoms could be predicted for that depth with an empirical formula; thus, the object contrast corresponding to bare detectability could also be computed.

Acknowledgments

Work was supported in part by an Industrial and Economic Development Grant from the University of Wisconsin—Madison Graduate School and a grant from Gammex, Middleton, Wisconsin, USA.

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APPENDIX I. TEST OF TOTAL INTERNAL REFLECTION PRODUCED WITH ALUMINA AND PLATE GLASS

To test the effectiveness of the plate glass and alumina reflectors in producing total internal reflection such that for sufficiently small angles of incidence (90° being perpendicular incidence), mode conversion to shear waves or Rayleigh waves does not occur (Lamkanfi et al. 2009), a phantom was constructed with parallel alumina and plate glass reflectors at opposite ends. The phantom is filled with tissue-mimicking background material (no spheres). The surfaces of the vertical plates are about 23 cm apart, and two scanning windows exist on the top of the phantom. One scanning window has a 1-cm ROC half-

cylindrical shape centered 5 cm from the plate glass reflector, and the other is flat and has the dimensions 5 × 10 cm with its center line 5 cm from the alumina reflector. The curved window provides for coupling of a 1-cm-ROC curved array with a sector angle of about 153°, and the flat window allows coupling of a phased array with a sector angle of about 90°. Thus, images can be created in which only one reflector is involved and the other side of the phantom consists entirely of tissue-mimicking material.

Note that the alumina reflector in the phantom described has a surface roughness of 6 μm, according to the manufacturer (CoorsTek, Golden, CO, USA).

With respect to the tissue-mimicking material in the phantom, the component materials are the same as in the sphere-containing phantoms except that the volume percentage of 3:1 ultra-filtered milk was reduced by a factor of 4/5 to yield a value of attenuation coefficient ÷ frequency ≈ 0.39 dB/cm/MHz instead of 0.5 dB/cm/MHz, so that the penetration depth was greater, ensuring availability of the data for analyses. Measurements were carried out at 22°C and 5 MHz using the procedure described by Madsen et al. (1999, see pp. 617 and 618, UWLMP section). The propagation speed is $c = 1539$ m/s, and the attenuation coefficient ÷ frequency = 0.39 dB/cm/MHz.

An average over 10 images using the phased array is illustrated in Figure A1. Two rectangular areas—10 mm wide—represent where pixel averages over 5-mm vertical depth increments were computed. The two rectangles have equal dimensions and are at the same positions vertically; however, one is just to the left of the vertical line of slight diffuse reflections arising at the surface of the plate glass (left side in the image), and both rectangles are displaced the same distance from the vertical axis of symmetry of the image. Thus, if total internal reflection were perfect and the imaging were perfect, then graphs of the pixel averages over 5-mm vertical depth increments versus vertical distance would be identical. Because there may be asymmetry of the phased array sensitivity, the two graphs may not be identical. Thus, the transducer was rotated 180° about a vertical axis, and another set of 10 images were obtained and averaged. With rotation of the transducer, the reflector is on the right side.

In Figure A2 are graphs of the 5-mm-deep mean pixel values for the left and right rectangles. Note that the pixel averages over 5-mm-deep rectangles are zero except between about 5 and 16 cm (vertical distances). In Figure A3 are the corresponding graphs after rotation of the transducer.

The percentage by which the mean pixel values resulting from reflections differ from the mean pixel values not involving reflections where asymmetry of transducer sensitivity is corrected can be derived from the data in Figures A2 and A3 using the equation

$$\frac{1}{2} \sum_{i=1}^2 \frac{R_i - N_i}{N_i} \times 100\% \quad (\text{A1})$$

where R_i and N_i are the mean pixel values on the reflector side and non-reflector side, respectively, and $i = 1$ corresponds to the reflector being on one side, and $i = 2$ corresponds to the reflector being on the other side. The results are summarized in Figure A4.

Figures A5–A7 correspond to Figures A1–A3 where the transducer is a 1-cm-ROC curved array and the reflector is alumina. The sector angle is about 153° . Total internal reflection fails for sector angles greater than about 138° corresponding to angles of incidence between 69° and 90° , as evidenced in the upper left side of the image in Figure A5 distal to the reflector.

APPENDIX II

Given that the sites involved should approximate a sphere with all sites lying inside a sphere of diameter D , the possibilities are (i) a single site; (ii) eight sites at the corners of a cube with side $D/4$; (iii) seven sites defining the center and corners of a tetrahedron (as employed); and (iv) nine sites, one at the center and eight at the corners of a cube with side $D/2$.

A single site (option 1) is unacceptable because random fluctuations in the background speckle would trigger an overwhelming number of false positives.

Option 2 is a possible choice with the greatest distance between two sites (at opposite corners of the cube) being $3^{1/2}D/4 \approx 0.43D$.

For option 3 (the one used in the software), the maximum distance between sites is $D/2$, slightly greater than for option 2.

The maximum distance between sites for option 4 is $3^{1/2}D/2 \approx 0.87D$. This is considered unacceptable when considering beam width effects in the lateral and elevational dimensions and pulse envelope effects in the axial dimension, as well as the statistical nature of the speckle pattern; thus, it is unlikely that all eight corner sites would be below the threshold (relative to the mean background level) for less detectable spheres.

Using option 3, we found that the number of spheres detected in any depth interval generally agrees with the mean number expected. Thus, almost all spheres detected by human observers in the image sets were detected by the software. Option 2 might be addressed in future work for comparison with option 3.

Fig. A1.

Average of 10 images obtained using a phased array where the plate glass reflector is on the left (*faint vertical line of diffuse reflections at the surface of the reflector*). Green rectangles indicate where mean pixel values were computed over 5-mm vertical increments.

Fig. A2.

Plot of the data with blue computed in the left rectangle in Figure A1 and red computed in the right rectangle. Pixel standard deviation bars are also illustrated. The abscissa is vertical depth.

Fig. A3.

Plot of the data when the reflector is on the right side with blue computed in the left rectangle and red computed in the right rectangle. The abscissa is vertical depth.

Fig. A4.

Percentage by which the mean pixel values resulting from reflections differ from the mean pixel values not involving reflections where asymmetry of transducer sensitivity is derived from the data in Figures A2 and A3 using eqn (A1). The abscissa is vertical depth.

Fig. A5.

Wide-sector (153°) 1-cm-radius-of-curvature transducer with alumina reflector on the left. Total internal reflection fails for sector angles greater than about 138° corresponding to angles of incidence between 69° and 90° , as evidenced in the upper left side of the image distal to the reflector.

Fig. A6.

Plot of the data with blue computed in the left rectangle in Figure A5 and red computed in the right rectangle. Pixel standard deviation bars are also illustrated. Note that for vertical depths from 0 through 1 cm, total internal reflection fails. The abscissa is vertical depth.

Fig. A7.

Plot of the data when the reflector is on the right side with blue computed in the left rectangle and red computed in the right rectangle. Again, for vertical depths from 0 through 1 cm, total internal reflection fails. The abscissa is vertical depth.

Fig. A8.

Percentage by which the mean pixel values resulting from reflections differ from the mean pixel values not involving reflections, where asymmetry of transducer sensitivity is derived from the data in Figures A6 and A7 using eqn (A1). Note that the value at 9.5 cm corresponds to a non-reflector mean pixel value of about 3; thus, although the right and left mean pixel values differ very slightly from one another, the percentage difference value is exaggerated. The abscissa is vertical depth.

Fig. 1. Human observer two-alternative-forced-choice results versus mean lesion signal-to-noise ratio (LSNR) values for 2-, 3- and 4-mm-sphere phantoms. Adapted from Kofler et al. (2005, Fig. 7b).

Fig. 2.

Photograph of the 3.2-mm-sphere phantom. The conical scanning window is directed upward, and there is a flat scanning window on the near end. The metallic appearance of the scanning windows results from their being formed from plastic-coated aluminum foil (PCAF) for suppression of desiccation of the water-filled tissue-mimicking material constituting the bulk of the phantom volume. The white covering on the remainder of the phantom is another form of PCAF for desiccation suppression.

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Fig. 3.

End-view diagram of the 3.2-mm-sphere phantom illustrating the plate glass reflectors and randomly distributed low-echo spheres. TM = tissue-mimicking.

Fig. 4. Photograph of the 2-mm-sphere phantom illustrating the conical scanning window which accommodates convex arrays with radii of curvature from 0.5 through 3 cm. A flat scanning window is on the opposite side. This phantom is for use at higher frequencies (7–15 MHz).

Fig. 5. End-view and top-view diagrams of the 2-mm-sphere phantom. This prototype phantom was produced with one plate glass reflector and one parallel alumina (Al_2O_3) plate reflector. The alumina reflector has a very high density and acoustic propagation speed, providing total internal reflection for angles of incidence as small as 16° , compared with about 30° for glass.

Fig. 6. Two-part mold for producing 1044 3.2-mm-diameter low-echo spheres. Holes and opposing brass pegs in the corners ensure alignment of the hemispheres to form spheres.

Fig. 7. Apparatus for acquisition of image sets. (a) View from side not showing stepper motor. (b) View showing stepper motor.

Fig. 8.
 Depiction of the square array of centers of overlapping square areas A over which mean pixel values are computed. The areas A have sides $= 2D/3$, and the centers are $D/4$ from their nearest neighbors.

Fig. 9.

Image of a 4-mm-sphere phantom made with a convex array. One of the plane reflectors (on the left) is alumina instead of plate glass, and a vertical line of elevated echoes occurs at the slightly rough surface of that alumina plate. There is a parallel 3-mm-thick plate glass reflector 10 cm to the right of the alumina plate. The maximum width of the sector image is 18.6 cm.

Fig. 10.

(a) Cropped image from Figure 9 with the vertical line of low-level diffuse echoes at the flat alumina plate surface removed. (b) Mapping of mean pixel values in gray scale. (c) Sets of seven sites identified with a sphere center in Figure 9. (d) Depth intervals distinguished with different gray levels. Note that some spheres are detectable in Figure 9 and (a) and (b) of this figure, but no evidence of them is seen in (c). This occurs when either no centers of sets of seven exist in Figure 9 or there are too few (*e.g.*, 2 or 3) to be evident in (c).

Fig. 11.

Left: Mean lesion signal-to-noise ratio (LSNR) values as a function of depth as determined with two independent sets of 25 images corresponding to Figures 9 and 10. Right: Number of spheres detected in each 5-mm depth interval. Reproducibility is demonstrated, but use of all 50 images in the set is recommended. See Figure 12.

Fig. 12.

Left: Mean lesion signal-to-noise ratio values as a function of depth using all 50 images corresponding to those in Figure 11. Right: Number of spheres in each depth interval. The curve fit looks good, implying that the number of spheres in each 5-mm depth interval in the focus range (about 25) is adequate.

Fig. 13.
One of 80 parallel linear array images of the 4-mm phantom with a focus at 3 cm.

Fig. 14.

Three successive images of the 80 separated by $D/4 = 1$ mm. The green x's identify the determined centers of the spheres identified with the respective images. Where there are no x's, the sphere center was identified with a nearby image.

Fig. 15.

Results for the 4-cm-wide 3-cm-focus linear array addressed in Figures 13 and 14. The image set 1–40 is independent of the image set 41–80. The agreement for the two sets is reasonable, but should be better. Note that the mean number of spheres in a depth interval is about 7. LSNR = lesion signal-to-noise ratio.

Fig. 16.

Left: Results for the 4-cm-wide 3-cm focus linear array addressed in Figures 13–15 using all 80 image frames corresponding to Figure 15. Right: Number of spheres detected in each depth interval. The curve fit is much better for this doubling of the number of spheres per centimeter of depth interval. Doubling the number of detected spheres in the 3-cm focus zone would better define the minimum (most negative) value of the mean lesion signal-to-noise ratio (LSNR), but the limited area of the window in this phantom precluded obtaining more than 80 images.

Fig. 17.
An image using the 3.2-mm-diameter sphere phantom and a 4-MHz linear array focused at 3 cm.

Fig. 18.

Reproducibility result for two independent sets of 70 data images with a mean number of sphere centers is about 13 per 5-mm depth interval in the focus range. Reproducibility is demonstrated. LSNR = lesion signal-to-noise ratio.

Fig. 19.

Result using both sets of 70 independent data images corresponding to Figure 18. The number spheres per 5 mm depth interval in the focal depth is about 25 corresponding to the recommended, and the curve-fit is very good. LSNR = lesion signal-to-noise ratio.

Fig. 20.
Sector image (convex array) at 4.5 MHz with multiple focuses at 4-, 8- and 12-cm depths.
The spheres are 3.2 mm in diameter.

Fig. 21.

Reproducibility results for a multiple lateral focus (4, 8 and 12 cm) case corresponding to Figure 20. Reproducibility is excellent at a mean number of sphere centers per 5-mm depth interval of 25. The lateral focusing at 8 and 12 cm is barely evident, focusing presumably being strongly dominated by fixed elevational focusing at a depth of about 4 cm. LSNR = lesion signal-to-noise ratio.

Fig. 22.

Reproducibility results for the case corresponding to Figure 20, except that there is a single focus at a depth of 10 cm. Reproducibility is excellent at a mean number of sphere centers per 5-mm depth interval of 25. The lateral focusing at 10 cm is not evident, focusing presumably being strongly dominated by fixed elevational focusing at a depth of about 4 cm. Note that there is a distinction between these results and the multiple focusing case in Figure 21, that is, a shallower minimum at about -7.9 instead of about -8.7 for the multiple-focus case. LSNR = lesion signal-to-noise ratio.

Fig. 23.

Reproducibility results for the case corresponding to Figure 20, except that there is a single focus at a depth of 4 cm. Reproducibility is excellent at a mean number of sphere centers per 5-mm depth interval of 25. The 10- and 4-cm-focus cases are almost indistinguishable, the 4-mm case having a slightly deeper minimum (negative value). LSNR = lesion signal-to-noise ratio.

Fig. 24.

Image of the 2-mm-sphere phantom made with a 1.5-cm radius of curvature curved array with lateral focus at 3 cm.

Fig. 25.

Reproducibility results corresponding to Figure 24. Images 1–50 are independent of images 51–100. Although reproducibility is reasonable, the plotted points are spread more than desired and error bars are rather large. Note that the number of spheres detected per depth interval in the focus range is about 15, which is less than the minimum of 25 recommended. LSNR = lesion signal-to-noise ratio.

Fig. 26.

Result using all 100 images in the image set that gave rise to Figure 25. The number of spheres per depth interval in the focus region (depth range: 1–1.5 cm) is about 30. The number of spheres detected per depth interval in the depth range 2.5–3.5 cm is <20, compromising the accuracy in that range. LSNR = lesion signal-to-noise ratio.

Fig. 27.

Image of the 2-mm-sphere phantom made with a high-frequency linear array (operating at 15 MHz) (laterally) focused at 4 cm.

Fig. 28.

Reproducibility results corresponding to Figure 27. Images 1–100 are independent of images 101–200. Reproducibility is demonstrated with about 35 detected spheres per depth interval in the focal depth range, which again must be at the elevational focus of about 1.5 cm.

Fig. 29.

Result using all 200 images in the image set that gave rise to Figure 28. The number of spheres per depth interval in the focus region (depth range: 1–1.5 cm) is about 70. LSNR = lesion signal-to-noise ratio.

Fig. 30.

Mean lesion signal-to-noise ratio (LSNR) values versus depth for ordinary B-mode images generated at nominal frequency 8 MHz with the Siemens 2000S and transducer array 18L6.

Fig. 31.

Mean lesion signal-to-noise ratio (LSNR) values versus depth for ordinary B-mode images generated at nominal frequency 10 MHz with the Siemens 2000S and transducer array 18L6. Except for the nominal frequency, all console settings were the same as in the 8-MHz case (Fig. 30).

Fig. 32.

Mean lesion signal-to-noise ratio (LSNR) values versus depth for ordinary B-mode images generated at nominal frequency 15 MHz with the Siemens 2000S and transducer array 18L6. Except for the nominal frequency, all console settings were the same as in the 8- and 10-MHz cases (Figs. 30 and 31).

Fig. 33.

Mean lesion signal-to-noise ratio (LSNR) values versus depth for images generated at nominal frequency 15 MHz with the Siemens 2000S and transducer array 18L6; however, spatial compounding (SieClear 5 setting) was used instead of ordinary B-mode images. Except for the spatial compounding, all console settings were the same as the case in Figure 32.

Fig. 34.

Mean lesion signal-to-noise ratio (LSNR) values versus depth for images generated at nominal frequency 15 MHz with the Siemens 2000S and transducer array 18L6; however, tissue harmonic imaging was used instead of ordinary B-mode images. Except for the spatial compounding, all console settings were the same as in the case in Figure 32.

Fig. 35.

Results for a Siemens S2000 and 7CF2 3-D (swept convex array) transducer focused at 4 cm and operated at 4.5 MHz in 2-D mode.

Fig. 36.

Results for an Ultrasonix “Sonix Touch” scanner with a 4DC7-3 3-D (convex array) transducer, operated at 4 MHz in 2-D mode and focused at 4 cm. The sector angle and all other console settings mimicked those for the Siemens case (Fig. 35).

Table 1

Propagation speeds and (water-corrected) attenuation coefficients of the tissue-mimicking materials in the phantoms at 22°C*

Tissue-mimicking material	Speed (m/s)	α_0 (dB cm ⁻¹ MHz ⁻ⁿ)	n
4-mm-diameter phantom background	1542	0.387	1.22
4-mm-diameter phantom spheres	1543	0.384	1.17
3.2-mm-diameter phantom background	1542	0.406	1.09
3.2-mm-diameter phantom spheres	1542	0.416	1.16
2-mm-diameter phantom background	1536	0.303	1.21
2-mm-diameter phantom spheres	1538	0.295	1.19

* The results are given in curve-fitted form assuming the exponential form $\alpha = \alpha_0 f^n$, where α is the attenuation coefficient, f is the frequency and α_0 and n are fitting constants. For the lower-frequency phantoms (4- and 3.2-mm phantoms), the frequency range is 2.5–7.5 MHz, and for the higher-frequency phantom (2-mm phantom), the frequency range is 5–15 MHz.