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Evaluation of large aperture imaging through the *ex vivo* human abdominal wall

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Abstract

Current clinical abdominal imaging arrays are designed to maximize angular field of view rather than the extent of the coherent aperture. We demonstrate in *ex vivo* experiments the use of a large effective aperture to perform high-resolution imaging even in the presence of abdominal wall-induced acoustic clutter and aberration. Point and lesion phantom targets were imaged through a water path and through three excised cadaver abdominal walls to create different clinically relevant clutter effects with matched imaging targets. A 7.36 cm effective aperture was used to image the targets at a depth of 6.4 cm and image quality metrics were measured over a range of aperture sizes using synthetic aperture techniques. In all three cases, although degradation compared to the control was observed, lateral resolution improved with increasing aperture size without loss of contrast. Spatial compounding of the large aperture data drastically improved lesion detectability and produced contrast-to-noise ratio improvements of 83%–106% compared to the large coherent aperture. These studies demonstrate the need for the development of large arrays for high-resolution abdominal diagnostic imaging.

Keywords

Large aperture; Low f-number; Swept aperture; High resolution; Synthetic aperture; Abdominal imaging; Image quality; Acoustic clutter

Introduction

The prevalence of obesity is a serious problem for ultrasound imaging. Nearly 35% of adults in the United States were considered obese as of 2012, a rate that has remained fairly constant over the past decade (Ogden et al., 2014). Obesity is tied to an increased rate of diabetes, the combination of which is a risk factor for a number of different cancers, heightening the need for screening tools in the obese population. For instance, obese patients with diabetes suffer from 2.5 times higher risk of hepatocellular carcinoma than the general

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population (Klysiak et al., 2014). Typical screening for hepatocellular carcinoma using ultrasound relies on identifying subtle differences in hypoechoic or isoechoic structures in the liver that are easily missed in the presence of acoustic clutter or poor resolution, especially in a cirrhotic liver (Irshad et al., 2012; Virmani et al., 2013).

Transabdominal fetal imaging is particularly difficult in obese women due to the naturally added imaging depth [Tsai et al., 2015]. Obesity is a risk factor for a number of both maternal and fetal complications that must be carefully monitored throughout the pregnancy. Fetal cardiac imaging presents the largest challenge, with obesity nearly doubling the rate of suboptimal visualization, increasing from 18.7% to 37.3% for normal versus obese patients in diagnostic scans (Hendler et al., 2004). Typical first- and second-trimester scans for cardiac defects include identification of the four-chamber view, Doppler imaging of inflow and outflow, and various measurements of valve and vessel sizes (Becker and Wegner, 2006; DeVore and Medearis, 1993). Skeletal features must be identified throughout development, including measurement of the head, limbs, bladder, stomach and umbilicus (Becker and Wegner, 2006). The size of the nuchal translucency during the first trimester has been identified as a significant marker for abnormal development, requiring precise differentiation around the cutoff of 3.5 mm (Souka et al., 2001).

Despite advancements in transducer construction, pulse sequencing, and signal processing, in difficult clinical situations there remain key impediments to diagnostic ultrasound imaging such as penetration, resolution, and clutter. Fundamentally, a longer array made up of many individual elements would produce images with improved lateral resolution and penetration. Preliminary phantom and *in vivo* work suggests that increasing aperture size improves target detectability at least up to the extent of current commercial arrays [Bottenus et al., 2013]. However, linear system theory is insufficient to predict the performance of such an array for *in vivo* imaging. Waves emitted from and received in different sections of the array experience unique realizations of the clutter-generating body wall. Reverberation clutter superimposes spatially incoherent noise on the received data [Pinton et al., 2014]. Aberration distorts the point spread function (PSF), broadening the main lobe and increasing off-axis scattering [Moshfeghi and Waag, 1988].

This work seeks to evaluate in a controlled, clinically relevant context the role of increased aperture size in improving image quality in the presence of acoustic clutter. Pulse-echo imaging using a large synthetic aperture setup was performed through three excised cadaveric abdominal walls that represented a range of expected clutter levels. We observed improved lateral resolution and target detectability with increasing aperture size despite the effects of the abdominal wall. Although contrast and contrast-to-noise ratio remained fairly constant with aperture size, both metrics were improved by using spatial compounding to take advantage of the large available aperture extent. These studies may encourage the future development of large aperture imaging systems for clinical abdominal imaging tasks.

Materials and Methods

The feasibility of evaluating large aperture imaging configurations with existing clinical hardware and the swept synthetic aperture method has been previously established [Bottenus

et al., 2016; *Zhang et al.*, 2016]. The overall goal of the experimental setup was to provide matched target images for each of the abdominal wall samples to evaluate the impact of clutter at varying aperture sizes. The swept synthetic aperture method was used to produce as large an effective aperture as the available equipment allowed – a 6.4 cm sweep equivalent to a 7.36 cm aperture – and to synthetically create images from a range of aperture sizes.

System overview

An overview of the electronics in the experimental setup and their communications is given in Fig. 1. The Siemens SC2000 ultrasound scanner (Siemens Healthcare, Mountain View, CA) was used with custom pulse sequences, described below, to acquire I/Q ultrasound data at a sampling rate of 2.5 MHz. The transducer used was the Siemens 4Z1c volumetric (2-D) array, described in Table 1. Sub-apertures of 3×3 elements summed without phasing were used in receive to reduce the channel count. A 3-D mesh model of the transducer produced using the Faro ScanArm laser scanner (Faro, Lake Mary, FL) was used to create a rigid custom attachment to the translation stage using Soft PLA plastic (MatterHackers, Foothill Ranch, CA) with the Lulzbot Taz 5 3-D printer (Aleph Objects, Inc., Loveland, CO).

The translation stage was run by the Newport XPS-Q8 motion controller (Newport, Irvine, CA). The setup consisted of three Newport UTM100/ILS100 linear translation stages ($5 \mu\text{m}$ on-axis accuracy and $1.5 \mu\text{m}$ unidirectional repeatability) mounted in orthogonal directions and one Newport URS50 rotation stage (0.03 degree accuracy and 0.002 degree unidirectional repeatability) mounted on the last translation stage axis to rotate in the x-z plane. The translation stages had a maximum travel distance of 10 cm in all directions. The transducer was mounted with the face approximately 3.8 cm from the point of rotation, reducing the available lateral extent for the swept arc.

The laptop PC acted as a controller for the other two devices, synchronizing movement and data acquisition using a Python program. For each acquisition, the program read a parameter file containing the appropriate scan geometry and scanner configuration. The SC2000 scanner was set up with the appropriate focal configuration and output settings over the local network. For each motion step, the motion controller was sent a command to move the translation stage to the desired (x, y, z, θ) position, representing its four degrees of freedom (three translation, one rotation). After a 100 ms settling time, the position was read from the stage and logged to a file. At each position, a command to acquire a single frame of I/Q data, as defined below, was issued to the SC2000. After all motion steps were complete, the program moved the data from the scanner to local disk with the log of positions and restored the normal imaging mode.

Pulse sequencing

Three types of pulse sequences were used during the experiment. The first was a full synthetic aperture (FSA) sequence [*Karaman and O'Donnell*, 1995] consisting of 192 transmit events and 192 receiving channels. A wave was emitted from each 3×3 sub-aperture independently with zero delays and all receiving channels (corresponding to the sum over 3×3 element groups) were recorded for each emission. The wave from this small

aperture is effectively a diverging wave at the target depth. This sequence was used as a baseline for array performance, providing a point of comparison to the current clinical implementation. The total time for this sequence was approximately 1 minute.

The second type of sequence was the swept synthetic aperture (SSA) sequence [Bottenus et al., 2016]. Data from 192 receive channels were recorded for a frame of data at each position in the sweep. Each frame consisted of twelve 3×3 element emissions, one on each of the laterally centered sub-apertures through the elevation dimension. This sequence allowed reconstruction of data with the full elevation aperture in transmit. A sweep consisted of 207 steps across an arc of 6.4 cm with a 6.4 cm radius ($\lambda/2$ steps), always pointing the transducer toward the center of the arc. The total time for this sequence was approximately 18 minutes.

The final type of sequence was used to calibrate the speed of sound in the water tank. The twelve-transmission sequence from above was performed at ten axial depths from a point target, stepping in increments of 1 mm.

Phantom targets

Annotated photographs of the water tank and phantom setup are shown in Fig. 2. Two different phantom targets were used to characterize resolution and contrast in the experiment. A custom point target phantom was produced by the University of Wisconsin-Madison (Madison, WI). The phantom was made of low-scattering agarose with $200 \mu\text{m}$ glass beads sparsely scattered within to act as isolated point targets. The phantom measured 10.5 cm high, 13 cm wide, and 10.5 cm thick. The phantom consisted of a 13 cm diameter cylinder section on top of a rectangular base. This curved surface was intended to reduce the angle of incidence during the transducer sweep. A stainless steel plate was embedded in the phantom's base for magnetic mounting to the tank.

An anechoic lesion phantom was created in two steps. First, a tapered cylindrical gelatin phantom (diameter 6.3 cm to 7.5 cm) was created by mixing 265 mL deionized water with 14 mL n-propanol and 18 g type A gelatin until dissolved. The mixture was placed in a vacuum chamber for five minutes, then held at 45 C in a water bath for one hour. Afterwards, 45 g graphite was added with gentle mixing. Once the mixture cooled to 33 C, a formaldehyde dilution of 20 mL deionized water and 3 mL formaldehyde was added and mixed gently. The phantom was allowed to set overnight while rotating the mold to suspend the graphite [Madsen et al., 1978]. The second step was to use a 9 mm diameter cork bore to create a hole through the axis of the phantom and to fill the hole with a similar gelatin mixture without graphite (anechoic). Due to the elasticity of the phantom, the resulting hole was smaller than the bore size. The phantom was pinned to two custom C-shaped brackets at its ends and attached to a steel plate for magnetic mounting to the tank. In Fig. 2, the cylinder runs perpendicular to the page, toward the front and back of the tank, such that a left/right scan shows a circular cross-section. The phantom was held such that the cylinder sloped from front to back to avoid specular reflection at its boundaries.

The phantoms were placed in a 45 g/L saline solution to achieve a sound speed of 1540 m/s in the water tank at 23 C [Lovett, 1978]. Both phantoms were held such that the target structure was approximately 6.4 cm from the transducer face.

Tissue preparation

Sections of the anterior abdominal wall were excised from three cadaver specimens. Sample 1 was a 73 year old caucasian male, 5' 10", 160 pounds (cause of death: metastatic pancreatic cancer). Sample 2 was a 72 year old caucasian male, 6' 0", 167 pounds (cause of death: lung cancer). Sample 3 was a 48 year old black male, 5' 10", 218 pounds (cause of death: stroke). Samples measured at least 10 cm square and consisted of hair, skin, subcutaneous fat, and some muscle. The measurements of these tissue layers before mounting are given in Table 2. Sample 2 had only a partial muscle layer, varying in thickness across the sample. In one study of body wall thickness, the median abdominal thickness (skin and subcutaneous tissues) of 23 randomly selected adult males ranged from 1.5 to 3.5 cm depending on the position within the abdomen [Sandler et al., 2010]. The three samples studied fall on the upper end of that range, representing potentially difficult imaging cases.

Samples were frozen for transport and thawed under refrigeration for 24 hours before mounting. Samples were lightly stretched across a custom tissue holder with a 6.4 cm radius arc in the lateral dimension to match the curvature of the transducer sweep. Samples were pinned to the frame with straight pins and the points were capped with putty for safety. The holder and the three samples are shown in Fig. 3. Samples were stored separately in 45 g/L saline solution for several hours and manually agitated to remove air bubbles from between the tissue layers. This time duration was not expected to affect the attenuation properties of the tissue, but may have a small effect on the overall magnitude of backscatter [Bamber et al., 1977].

Experimental protocol

The phantoms were fixed in the tank for the duration of the experiment to maintain registration between the different trials. The translation stage positioned the transducer in elevation over either the point or lesion target for a given trial, providing repeatable views of the phantoms despite moving the transducer. The control experiment was performed first without tissue present. The complete sequence consisted of an FSA acquisition, an SSA acquisition, approximately 60 minutes of additional experiments not described in this work, and a final FSA acquisition to check for consistency during the 80 minute scan duration.

After the control scan, a tissue sample was transferred from its saline bath into the water tank and clipped onto the slide rails. This allowed the tissue to be positioned over either phantom without movement of the phantoms or tank. Both targets were scanned as before. The tissue sample was manually positioned over each target, so the two targets likely corresponded to different elevation realizations of the tissue. The samples were positioned to sit immediately over the phantom, coupled to the transducer by saline. Samples 2 and 3 extended to the transducer surface due to their thickness. Tissue samples spent a minimum of 160 minutes in the scanning tank, not accounting for some additional preparation time by

the operator. No changes to the tissue structure were apparent during this time for any of the tissue samples.

After all the tissue samples were scanned, a final control scan was performed (FSA and SSA) to verify registration of the phantom, tank, and transducer.

Calibration

Two types of calibration were performed. First, speed of sound $\hat{c}$ was estimated using a linear regression of induced axial motion Δz versus observed target depth z_{obs} :

$$z_{obs} = \left(\frac{c}{\hat{c}} - 1 \right) \Delta z + \hat{z}_0 \quad (1)$$

The position z_{obs} includes the changing position of the array relative to the target such that under an ideal calibration no motion would be observed with changing axial position. Ten frames of channel data were beamformed as described below, and the beamformed data from the twelve emissions from each position were summed to improve SNR. Each observed point location was estimated using the peak of the envelope image. An initial sound speed guess of $c = 1500$ was used, although the result is fairly insensitive to the initialization. The estimated intercept $\hat{z}_0$ was ignored.

Second, mechanical calibration was performed using the diverging wave sequence data. The data from each position were beamformed as described below and the N observed point locations $\vec{p}_i = [x_i, y_i, z_i]$ defined by the peaks of the envelope images in 3-D space were recorded. The 4×4 calibration matrix $\mathbf{X}$, a standard 3-D transformation matrix, was empirically refined to minimize the reconstruction precision (RP) of the points relative to the mean position $\bar{p}$, especially in the axial and lateral dimensions:

$$RP = \left(\frac{1}{N} \sum_{i=1}^N \|\vec{p}_i - \bar{p}\|^2 \right)^{1/2} \quad (2)$$

RP was calculated for all three dimensions independently, although it could alternatively be calculated for all three simultaneously to minimize the overall spread [Zhang et al., 2016].

Data processing

Data were processed offline for all sequences. The channel data were beamformed using interpolation based on the time-of-flight t_{ij} of the wave corresponding to an emission i to a point in the field $\vec{r}_0 = [x_0, y_0, z_0]$ and back to each receive channel j [Karaman and O'Donnell, 1995]:

$$s_{foc}[\vec{r}_0] = \sum_i \sum_j s[t_{ij}, i, j] \quad (3)$$

Although 2-D images were produced to more directly study the effect of growing the lateral aperture size, full 3-D focusing was applied to provide optimal image quality at all depths.

The diverging sources in the FSA sequence and the diverging SSA sequence appear as point sources at elements along the array. For a point source located at $\vec{r}_i = [x_i, y_i, z_i]$, an imaging point to be reconstructed $\vec{r}_0$, and a receive element located at $\vec{r}_j = [x_j, y_j, z_j]$, the necessary time delay is:

$$t_{ij} = \frac{\|\vec{r}_0 - \vec{r}_i\| + \|\vec{r}_j - \vec{r}_0\|}{c} \quad (4)$$

For the FSA scans, no mechanical calibration was required. For the SSA scans, the array element position vectors $\vec{r}$ were spatially transformed prior to beamforming using the transformation induced by the translation stage **B** and the calibration matrix **X**. The induced transformation is based on the 3-D translation **T** and 1-D rotation about the y-axis **R_y**:

$$\mathbf{B}(i) = \mathbf{T}(i) \mathbf{R}_y(i) \quad (5)$$

For a position vector to be transformed $\vec{r}$:

$$\vec{r}_{trans}(i) = \mathbf{X}^{-1} \mathbf{B}(i) \mathbf{X} \vec{r} \quad (6)$$

Before beamforming, the demodulated I/Q data were upsampled, remodulated with the appropriate carrier frequency, and converted to real RF data signals.

The synthetic aperture data were processed in several different combinations to study image quality. The FSA results provided a baseline for image quality with and without the tissue clutter. The FSA results were also used to adjust the speed of sound used for beamforming in the presence of tissue because the subcutaneous fat is expected to have a lower speed of sound than the saline or phantom. Since the images were only beamformed over a relatively small depth range, a correction was made to adjust the gross sound speed such that the point target appeared at the same depth as in the control.

The SSA results were studied using synthesized subarrays from positions across the array to examine variation in the clutter produced across the extent of the tissue samples. The transmit and receive apertures were synthetically grown by coherently combining data from successive array positions with a flat apodization. The effective subaperture length was reported as the average of the lengths of the synthesized transmit and receive arrays, or the sweep length plus half the receive array extent. Data from one position were omitted from the tissue 2 case due to scanner malfunction.

For each case, lateral resolution was measured using the envelope signal to find the full-width at half-maximum (FWHM). For the FSA data sets, the impact of clutter was measured by computing the ratio of the point target brightness to the mean speckle background level

and normalized by the control case [Nock et al., 1989]. The point brightness measurement was also performed for synthesized subarrays across the sweep to study the spatial variability of clutter. Lesion detectability was measured from the envelope signal using the contrast and contrast-to-noise ratio (CNR) for regions inside and outside the lesion in the 2-D axial-lateral plane. Contrast was computed as $20\log_{10}(\mu_i/\mu_o)$ and CNR [Smith et al., 1983] was computed as $(\mu_o - \mu_i) / \sqrt{\sigma_o^2 + \sigma_i^2}$.

Spatial compounding [Trahey et al., 1986] of the swept aperture data was performed by forming synthetic transmit subapertures as described above with half-aperture overlap (as recommended in O'Donnell and Silverstein [1987]), coherently combining all receive channel data corresponding to those transmit events, and incoherently averaging the envelope signal from all subapertures. In addition to contrast and CNR, speckle signal-to-noise ratio (sSNR) was computed based on the non-compressed envelope signal as μ_o/σ_o . Spatial decorrelation of images was measured by comparing the background speckle regions of interest of the non-compressed envelope between selected subaperture images using normalized cross correlation as a function of aperture translation. All pairs of subapertures were compared for a fixed effective aperture size and the resulting correlations were averaged by separation distance.

The dynamic range of displayed images was selected to provide a fair visual comparison [Bottenus and Ustuner, 2015]. The mean and variance in the identified speckle region of the log-compressed images were matched to the control case, highlighting differences in the point or lesion target. All image metrics were computed before this non-linear transformation was applied.

Results

System calibration

Sound speed calibration was performed using the control point target data. The calibration assumes a constant sound speed between the transducer and the point target. Fig. 4(a) shows the estimated sound speed based on the linear regression of induced motion versus observed point target depth. A sound speed of 1541.4 m/s was estimated with an R^2 value of 0.995. This value was used for beamforming the control data and as a reference for adjusting the effective sound speed in the tissue cases as described below. Residual error may come from uncertainty in the translation stage motion, orientation error (i.e. the axial motion is not aligned with the axial dimension of the transducer), or peak localization error due to the finite size of the point target (i.e. its apparent peak may change with changing insonification).

Manual calibration of the 6 degree of freedom transformation matrix that represents the relationship between final joint of the translation stage (the x-z rotation) and the face of the array was performed using the control point target data. The reconstruction precision was insensitive to some corrections that do not affect the relative position throughout the sweep such as elevation offset or y-axis rotation. Fig. 4(b) shows the axial-elevation spread of reconstructed points relative to the average point location. The axial RP was 13.9 μm , or

2.25% of the axial wavelength. The lateral RP was $41.4\ \mu\text{m}$, or 6.71% of the lateral resolution. No clear drift due to miscalibration or experimental error across the 207 positions was observed in these dimensions. Fig. 4(c) shows the axial-elevation spread of reconstructed points relative to the average point location. The elevation RP was $273.7\ \mu\text{m}$, or 10.0% of the elevation point spread function. This dimension showed a drift in elevation reconstruction position with transducer position. For the purposes of this experiment, the RP was sufficient without correction of this error.

Conventional aperture reference

The full synthetic aperture results provide a baseline for image quality with and without the tissue layers present. Fig. 5 shows the two phantom targets imaged in the control case and through the three different tissue layers. Regions of interest (ROIs) for quantifying image quality are shown in Fig. 6 for both phantoms. The point target was described using its lateral PSF, taken from the depth at which the maximum envelope value occurs (the white dashed line in Fig. 6(a)). The lesion image ROIs shown in Fig. 6(b) omit a 40 degree region in the lower left of the background and a 1.1 mm radius circle in the lesion to avoid a reverberant artifact and gas bubble respectively. Sound speeds of 1541.4 m/s, 1510 m/s and 1510 m/s were used to process tissue samples 1–3 respectively to place the point target at the correct depth.

The point target in the control case, Fig. 5(a), had a lateral FWHM resolution of 1.92 mm and was visible on top of the speckle background that had an average amplitude of $-42.1\ \text{dB}$ compared to the point. The lesion in Fig. 5(e) was visible with a contrast of $-22.78\ \text{dB}$ and a CNR of 1.67. It was easily identifiable from the phantom background but had poorly defined edges, making a border difficult to delineate. Fig. 7 summarizes the computed image metrics for both the control and tissue cases.

In all cases, little variation was observed in the lateral resolution of the point target. The maximum change in the FWHM, 7.3%, occurred in the tissue 3 case. The point target appeared translated laterally when imaging through tissue 2, although the point target otherwise looked undistorted by aberration. The change in background brightness relative to the point target was apparent in all three tissue cases. The smallest change was observed in tissue 2 (7.44 dB) and the largest was observed in tissue 3 (20.17 dB). This places a noise floor on the visibility of targets, as evidenced by the lesion targets. The contrast of the lesion dropped significantly in all three cases, rendering it barely visible to visual inspection. Contrast was reduced by 13.89–15.86 dB. CNR showed a similar loss in lesion detectability, decreasing by between 37.1% and 47.3%. The lesion metrics showed the best performance for tissue 2 and comparably poor performance for tissues 1 and 3, although all were similar. While all image metrics showed the same trend in severity of image degradation between the three tissue samples, it should be reiterated that it is likely that different segments of the tissue sample were imaged with each phantom. It is possible that tissue 2 induced less clutter than the other two samples due to its limited muscle layer, even though the fat layer was thicker than that of tissue 1.

Variation in the level of background clutter was observed across the spatial extent of the tissue samples. To study these levels, synthetic apertures with varying center positions were

formed from the SSA data using an effective lateral aperture size of 1.92 cm to match the FSA case. Fig. 8 shows the point target brightness compared to the average speckle background level for the control and each of the tissue samples. The average values across the sweep were similar to the FSA cases in Fig. 7, within ± 2 dB. For tissue 3, the highest background clutter levels corresponded to no visual point target detectability while the most visible target locations showed up to 20.4 dB less clutter. Tissue 1 showed clutter preferentially on one side of the sweep, differing by up to 10.5 dB. Variability of 5.6 dB was present even in the control sweep, possibly due to variations in the phantom media or asymmetry in the reflectance of the point target.

Large aperture imaging

Image quality as a function of aperture size was studied for the control and tissue samples, growing from the center of the synthetic array outward. Fig. 9 shows the point targets for two effective aperture sizes: one equivalent to the FSA aperture extent and one corresponding to the full sweep. FWHM lateral resolution was measured for each effective aperture size and the results are shown in Fig. 10 for f-number varying from 3.5 to 0.8. Main lobe resolution significantly improved in all cases, even in the presence of the aberrating tissue layers. Resolution improved roughly linearly with f-number as predicted for imaging in the absence of tissue. Tissue sample 2 produced a discrete jump in measured resolution due to a side lobe increasing above -6 dB in amplitude with growing aperture, seen in Fig. 9(c). Despite this distortion, which will be discussed further in the following section, lateral resolution still improved over the FSA case. The FWHM values at the largest aperture sizes for the four cases were 0.42 mm, 0.48 mm, 1.10 mm, and 0.89 mm respectively. Unlike in the FSA case, where lateral resolution was largely unaffected by the tissue samples, the SSA case showed distinct differences across the tissue cases. The FSA-equivalent SSA images showed similar results to the control FSA image marked by the black dashed lines in Fig. 10, reaffirming the effective aperture size description used for the SSA case.

The lesion phantom was used to measure contrast and CNR over the range of synthetic aperture sizes. Fig. 11 shows the FSA-equivalent images (top row) and the full swept extent images (middle row) across the control and different tissue samples. A bright streak was visible to the lower left of the lesion in the SSA images, corresponding to a specular reflection artifact off another part of the experimental setup, and was omitted from image metric calculations. Qualitatively, the lesion was visible at both pictured aperture sizes across all three tissues. Due to the increased resolution, the boundaries of the circular lesion were much better defined at the large aperture size. Even in the control case, without added clutter, the smaller aperture was unable to resolve the shape of the target. Clutter produced by the tissue layer reduced the contrast of the lesion relative to the control, but did not significantly distort the boundaries.

The measured contrast and CNR values for the range of synthesized apertures are shown in Fig. 12. Little variation was measured across the aperture sizes, but a drop in the image quality metrics was observed at the largest aperture sizes compared to the FSA-equivalent. Contrast was reduced by up to 1.14 dB and CNR was lowered by up to 9.8% in the most extreme cases. For tissue sample 3, which showed the largest drop in CNR, this may be

partially explained by the severe clutter observed at the edges of the sweep in Fig. 8. Target detectability defined by Smith et al. (1983) is made up of both resolution (number of speckle cells in the ROI) and CNR. The improved resolution, ranging from 1.6 to 4.3 times better, therefore results in an improved overall target detectability at the large aperture size.

Spatial compounding

The goal of spatial compounding in ideal imaging conditions is to incoherently average the spatially varying speckle texture to improve the sSNR, in turn raising the CNR by reducing the variance of the texture. In more realistic imaging scenarios, the clutter is also expected to vary spatially, providing another target for spatial compounding. Spatial compounding requires sacrificing resolution by subdividing one or both apertures so optimization should be performed for a given imaging task. In this case, the lesion ROI extended to the observable boundaries of the lesion to penalize the loss of lateral edge resolution.

The *ex vivo* results are shown in the bottom row of Fig. 11 and the image quality metrics as a function of subaperture size are shown in Fig. 13. The improvements made over the SSA imaging case are summarized in Table 3. The control case showed drastic suppression of the speckle texture, increasing sSNR by 127.18% and CNR by 103.51% at the expense of 5.89 dB of contrast. Note that the dynamic range matching applied to the images attempts to equalize the background variance so that the improved sSNR appears visually as improved lesion contrast. Spatial compounding greatly improved image quality in all three tissue cases. The tissue 3 images had the least improvement, an 83.46% increase in CNR. The tissue 2 case had the most improvement, a 105.91% increase in CNR and only a 0.52 dB loss of contrast. The speckle SNR curves in Fig. 13(a) and speckle correlation curves in Fig. 14 show similar behavior with and without the abdominal tissue clutter.

Discussion

These experiments support the hypothesis that very large imaging arrays will markedly improve image quality in abdominal applications. Difficult abdominal imaging tasks require an array design that provides sufficient penetration for the chosen target depth, adequate resolution for the diagnostic task, and tolerance of acoustic clutter effects that degrade the backscattered echo signals. While signal degradation by the abdominal wall was observed as expected, the overall image quality improved with increasing aperture size to the maximum length tested, a 6.4 cm transmit array and 8.3 cm receive array. Assuming $\lambda/2$ element pitch, the effective matrix array size was 238×47 (although the array was not sampled this way in the experiment). Large apertures are also well suited for spatial compounding. A larger array means that higher resolution can be maintained while compounding relative to a conventional array. Additionally, because aberration correlation lengths are on the order of several millimeters [Hinkelman et al., 1994], a longer array provides more independent images and clutter realizations of a target, which lead to improved compounding performance. The results also show that the speckle behavior, upon which compounding depends, does not vary greatly with or without the abdominal layer clutter. The optimum subaperture sizes for compounding are all similar, so compounding *in vivo* would not need to be an adaptive process.

Large aperture imaging naturally has several physical drawbacks, such as hardware and software complexity as well as increased transducer size and weight. High channel count and array complexity are problems that have already largely been solved in commercial matrix arrays, while the operator ergonomics problem may be solved by additional fixtures to help support the probe in the clinic. Selecting a transducer curvature that allows for coherent beamforming of all the elements (accounting for angular sensitivity) while still being practical for various body shapes may be a challenging task. Keeping such a large device coupled well to the skin would add difficulty to clinical scanning despite its concave shape. The curvature of a concave array also limits the lateral field of view when the entire aperture is used because the edge elements are steered inward. However, this can be overcome by using a smaller active aperture that has the correct directivity for those sections of the image. A lateral field of view of at least the transducer footprint could be easily achieved with only small losses in lateral resolution at the edges. Large arrays are well-suited to subcostal applications such as abdominal and obstetric imaging, although the ability to tilt the transducer towards targets under the ribs may be limited. The added array size may not be useful for typical intercostal imaging applications due to ribs blocking large groups of elements and narrow acoustic windows.

One anomaly in the data is the imaging performance through tissue 2. It demonstrated the worst point target degradation at the full aperture size but the smallest drop in point target brightness, the largest contrast and the highest CNR. The synthesized channel data for the point target, produced by averaging the focused data across multiple transmit positions for each receive position, showed a smoothly varying axial drift across the sweep on the order of 1.5 wavelengths that was consistent through elevation. It is unclear whether this drift was a result of motion of the target (e.g. the transducer pressing the tissue into the phantom and displacing it) or aberration. The same pattern was not present in the lesion target data, but unfortunately this does not distinguish between the two effects as the tissue was moved between the scans and the lesion phantom was physically smaller than the point phantom. When the focusing is corrected for this low frequency drift, the point target coherence is restored, the double peak artifact observed in Fig. 9 disappears, and a lateral resolution of 0.43 mm is obtained. The data was presented without this correction in case the origin of the artifact was indeed aberration, although we believe it to be a target motion artifact.

The *ex vivo* experimental setup required several concessions that differ from a true *in vivo* large aperture experiment, but we believe that these do not compromise the conclusions drawn from these results. The data from the swept synthetic aperture represent a subset of the data available in a fully sampled large array. A large array with access to every transmit/receive element pair would have even higher signal-to-noise ratio and signal redundancy, further improving image quality. It is assumed that the majority of aberration and reverberation effects are produced by the abdominal wall, not the homogeneous liver tissue beyond [Hinkelman et al., 1994]. That assumption implies that imaging at a constant f-number deeper in the liver (i.e. with a larger aperture) would perform at least as well, if not better, as the propagation path length increases relative to the abdominal wall thickness. Also, calibration of the translation stage apparatus was sufficient for the resolution of the system, but such uncertainty would not be present in a true large aperture system. The use of 3×3 element groups reduces channel count at the expense of angular sensitivity, which was

overcome for the desired field of view using the concave aperture shape. A large aperture design could similarly use this strategy to reduce channel count as needed.

Conclusions

We have demonstrated the success of using a large effective array for imaging both point and lesion targets through the *ex vivo* human abdominal wall. In challenging abdominal imaging scenarios, resolution and image quality with current signal processing techniques are insufficient and transmit frequency cannot be raised due to increased attenuation. The image quality improvement demonstrated in this work may justify the development of larger arrays for specialized diagnostic use where high resolution is needed. However, these results do not address the importance of harmonic imaging to modern clinical ultrasound. Future study should evaluate the image quality benefits of combining resolution improvement from large apertures with the clutter reduction of harmonic imaging.

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Figure 1.

Acquisition system diagram. The controlling PC (center) performed two-way communication with the translation stage and the Siemens SC2000 ultrasound scanner. Matrix channel data were stored for each transmission at each of the 207 translation stage positions.

Figure 2.

Translation stage experimental setup. Components in the two views, front and top, are labeled with letters. (A) Anechoic lesion target phantom. (B) Point target phantom. (C) Tissue support rails. (D) Saline, sound speed $c \approx 1540$. (E) Volumetric transducer mounted to translation/rotation stage, covered with black plastic except for the face. (F) Tissue support structure, pictured positioned over the point target phantom. Note that final design shown in Fig. 4 allowed for wider placement of the support rails.

Figure 3.

(a) Tissue holder model for mounting the tissue samples with setup geometry. A 1 radian sweep was performed, giving a 6.4 cm arc length at 6.4 cm radius. (b,c,d) Views of each tissue sample from the front and side after mounting. The tissues are referenced by number – 1, 2, 3 – from left to right. The transducer scanned horizontally, left/right on the top view.

Figure 4.

(a) Sound speed calibration based on ten axial transducer positions. Linear fit provided a sound speed estimate of 1541.4 m/s. (b) Axial-lateral reconstruction error, showing point centers from across the 207 position sweep. (c) Axial-elevation reconstruction error. Note the difference in axis scales.

Figure 5.

Full synthetic aperture results. Point targets are shown in (a–d) and lesion targets are shown in (e–h). The control acquisition is shown in (a,e). The tissue samples are shown in (b,f), (c,g), and (d,h). All images are dynamic range matched to the control case and 60 dB dynamic range is shown.

Figure 6.

Regions of interest (ROIs) for image analysis. (a) Point target ROIs. Horizontal dashed line represents the peak point intensity depth for extracting the lateral PSF and FWHM. The area between the concentric circles represents the region where the background mean relative to the point target was measured. (b) The inner dashed circle represents the lesion interior ROI, and the area between the outer concentric circles represents the background ROI.

Figure 7.

Image metrics computed with the regions of interest indicated in Fig. 6. (a) Lateral point spread functions extracted for the four samples at the depth of peak point intensity. (b) FWHM measured from the lateral PSF. (c) Ratio of point target magnitude to the mean speckle background, compared to the control. (d) Lesion contrast relative to the background. (e) Lesion contrast-to-noise ratio.

Figure 8.

Ratio of point target magnitude to the mean speckle background, relative to the control acquisition, for synthesized apertures across the sweep with effective aperture size equal to the FSA. The dashed lines represent the average value.

Figure 9.

Point target images. Columns (a–d) correspond to the control and three tissue samples respectively. (top row) Swept synthetic aperture images corresponding to an effective aperture size of 1.92 cm (f-number 3.3), matching the FSA results of Fig. 5. (middle row) Swept synthetic aperture images at the largest aperture size, 7.36 cm (f-number 0.87). (bottom row) Lateral PSFs for the two displayed aperture sizes, dashed line for the smaller aperture and solid for the larger aperture. The lateral PSFs are shown normalized to each point target amplitude.

Figure 10. FWHM resolution as a function of the effective aperture size, expressed as the f-number. The FSA f-number and FWHM from Fig. 7 are marked as black dashed lines for reference.

Figure 11.

Lesion target images. Columns (a–d) correspond to the control and three tissue samples respectively. (top row) Swept synthetic aperture images corresponding to an effective aperture size of 1.92 cm (f-number 3.3), matching the FSA results of Fig. 5. (middle row) Swept synthetic aperture images at the largest aperture size, 7.36 cm (f-number 0.87). The bright reflection artifact to the lower left of the lesion was excluded from the image metric analysis. (bottom row) Spatially compounded images at the optimized subaperture size. The dynamic range of all images has been matched based on the speckle background.

Figure 12.

(a) Contrast and (b) CNR of the lesion phantom target as a function of the effective aperture size.

Figure 13.

(a) Speckle SNR, (b) Contrast, and (c) CNR for the spatially compounded lesion targets as a function of subaperture size. The large circular markers indicate the selected aperture size to optimize CNR.

Figure 14.

Speckle correlation as a function of aperture translation for an effective synthesized aperture length of 1.52 cm, marked as the dashed black line. The black dotted line indicates half the aperture length, which was the separation of apertures used for compounding in the control case.

Table 1

Siemens 4Z1c transducer geometry, transmit, and receive configurations

Property	Value	Unit
Geometry	2-D	-
Lateral pitch	0.4	mm
Lateral size	1.92	cm
Lateral channels	48	channels
Elevation pitch	0.4	mm
Elevation size	1.44	cm
Elevation channels	36	channels
Center frequency	2.5	MHz
Frac. bandwidth (approx.)	0.6	-
Sub-aperture size	3 × 3	channels
I/Q sampling rate	2.5	MHz

Table 2

Tissue sample layer thickness measurements

Sample	Skin (mm)	Fat (mm)	Muscle (mm)
1	3.5	18.0	9.0
2	2.0	24.0	0.0–10.0
3	2.0	28.0	19.0

Table 3

Image metrics for spatially compounded lesion targets

	Ap. size (cm)	Speckle SNR		Contrast (dB)		CNR	
		Full	Compounded	Full	Compounded	Full	Compounded
Control	1.52	1.77	4.03	-24.17	-18.28	1.66	3.38
Control	1.49	1.79	4.20	-9.76	-7.58	1.16	2.36
Control	1.63	1.77	3.73	-9.91	-9.39	1.15	2.37
Control	1.87	1.74	3.55	-7.91	-6.62	0.99	1.81