Paper presented at the SPIE Medical Imaging meeting, Ultrasonic Imaging and Signal Processing, 2002:

Multi-Element Synthetic Transmit Aperture Imaging using Temporal Encoding

Kim L. Gammelmark and Jørgen A. Jensen
Center for Fast Ultrasound Imaging,
Ørsted•DTU, Bldg. 348,
Technical University of Denmark,
DK-2800 Lyngby, Denmark

Send correspondence to Kim L. Gammelmark, E-mail:
klg@oersted.dtu.dk

Multi-Element Synthetic Transmit Aperture Imaging using Temporal Encoding

Kim L. Gammelmark and Jørgen A. Jensen
Center for Fast Ultrasound Imaging, Ørsted•DTU, Build. 348,
Technical University of Denmark, 2800 Kgs. Lyngby, Denmark

Abstract
A new method to increase the signal-to-noise-ratio (SNR) of synthetic transmit aperture (STA) imaging is investigated. The new approach is called temporally Encoded Multi-Element STA imaging (EMESTA). It utilizes multiple elements to emulate a single transmit element, and the conventional short excitation pulses are replaced by linear FM signals. Simulations using Field II and measurements are compared to linear array imaging. A theoretical analysis shows a possible improvement in SNR of 17 dB. Simulations are done using an 8.5 MHz linear array transducer with 128 elements. Spatial resolution results show better performance for EMESTA imaging after the linear array focus. Both methods have similar contrast performance. Measurements are performed using our experimental multi-channel ultrasound scanning system, RASMUS. The designed linear FM signal obtains temporal sidelobes below -55 dB, and SNR investigations show improvements of 4-12 dB. The depth performance is investigated using a multi-target phantom. Results show a 30 mm increase in penetration depth with improved spatial resolution. In conclusion, EMESTA imaging significantly increases the SNR of STA imaging, exceeding that of linear array imaging.

Keywords: Ultrasound, linear array imaging, synthetic transmit aperture imaging, multi-element defocusing, linear FM signals, temporal sidelobe reduction, Field II, experimental ultrasound scanning system.

1 INTRODUCTION
One of the main problems in ultrasound imaging is the tradeoff between penetration depth and spatial resolution. The resolution is improved as the frequency increases, but the attenuation in soft tissue also increases with increasing frequency. Increasing the peak pressure is not possible since the Food and Drug Administration (FDA) has introduced restrictions on the maximum intensities, which may occur in the body, to avoid damage to the tissue and pain to the patient. For the conventional imaging techniques like linear array imaging the peak amplitude of the short excitation pulses used can, therefore, not be increased infinitely to overcome the loss in penetration depth, when high frequency transducers are used. This is because the spatial-peak-temporal-peak intensity ($I_{sptp}$) is much higher than the spatial-peak-temporal-average intensity ($I_{spfa}$) for these systems due to the transmit focusing. For B-mode imaging, the $I_{sptp}$ limit is very often reached much earlier than the $I_{spfa}$ limit for these systems, which limits the amplitude of the excitation pulse. To increase the penetration depth, $I_{spfa}$ must be increased. This is done by increasing the length of the transmitted pulse. For this purpose the conventional pulse cannot be used, since the axial resolution is directly proportional to the pulse length (for short pulse systems). Thus, some type of temporal encoding is necessary. This is offered by the linear FM signal, which has been used in radars for decades, because of its compression ability. These signals have recently been applied in ultrasound scanners, and they have shown good performance in terms of increasing the penetration depth, while maintaining the spatial resolution. [1, 2, 3]

Generally, linear array imaging systems obtain good spatial resolution due to the application of dynamic receive focusing. The resolution can, however, be improved by increasing the number of transmit foci. This is often done in modern ultrasound scanners at the expense of a reduction in the frame rate proportional to the number of foci. One way to obtain high spatial resolution, while keeping the frame rate high, is to use the synthetic transmit aperture (STA) imaging
technique. The inherent nature of this technique makes it possible to generate images with dynamic transmit and receive focusing using only a few transmissions. Therefore, STA imaging offers very high frame rates, which makes it suitable for real-time 3-D volumetric imaging. A problem in STA imaging is, however, the low signal-to-noise-ratio (SNR), due to the application of a single transmit element\(^1\). This means that the penetration depth obtained by STA imaging is much lower than that obtained by linear array imaging, which significantly limits its clinical application. However, if the SNR can be increased to that of linear array imaging, images with better image quality can be generated, while keeping a high frame rate. For this purpose the linear FM signal can be applied with great advantage. Since no focusing is applied in transmit, \(I_{\text{SNR}}\) is no longer the main problem, and therefore the amplitude can be increased. Also, previous research has shown [4, 5, 6] that the application of multi-element subapertures in STA imaging to emulate a single element transmission yields good spatial resolution results and also increases the SNR proportional to the number of elements used in the subaperture. Therefore, the purpose of this study is to investigate the application of both these techniques in STA imaging and make a comparison to linear array imaging. Both simulations using Field II and measurements using our experimental multi-channel ultrasound scanner are used for the investigation.

## 2 SYNTHETIC TRANSMIT APERTURE IMAGING

In conventional STA imaging a single element is used at each transmission to produce a broad beam that interrogates a large region of the medium. The echoes resulting from scattering in the medium are recorded using all elements in the aperture, and the procedure is repeated until a desired number of element have been used for transmission. For each transmission the echoes received by each element in the aperture contain information about all scatterers in the interrogated region, and since no focusing has been applied on transmit, the receive aperture can be steered in any direction and focused at any point within this region. That is, a complete set of dynamically focused receive beams can be formed (simultaneously) in all directions in the region. These lines constitute a full image with low lateral resolution due to the single transmit element. The individual images are dynamically focused on transmit in a similar manner and subsequently summed to form the final high resolution image.

### 2.1 STA Beamforming

To describe STA beamforming in more detail, let \(r_{mn}(k)\) be the digitized signal received by element \(n\) when transmitting with element \(m\). Also, let \(\tau_n^l(k)\) be the dynamic focusing times for receive element \(n\) and line \(l\) in the low resolution image. Then, each line in the low resolution image for transmit element \(m\) is calculated by

\[
s^l_m(k) = \sum_{n=1}^{N} W_n(k) r_{mn}(k - \tau_n^l(k)f_s), \quad l = 1, 2, 3, \ldots, L
\]  

(1)

where \(W_n(k)\) is the dynamic apodization values for receive element \(n\), \(N\) is the number of receive channels, and \(L\) is the number of lines in the image. This is a formulation of the conventional delay-and-sum beamformer. If the product \(\tau_n^l(k)f_s\) is not an integer, interpolation is needed to find the correct value.

After the low resolution images for all \(M\) transmit elements have been beamformed, the final high resolution image is created by delaying and summing the corresponding lines from each low resolution image. In particular,

\[
S^l(k) = \sum_{m=1}^{M} W_m(k) s^l_m(k - \tau_m^l(k)f_s)
\]

\[
= \sum_{m=1}^{M} W_m(k) \sum_{n=1}^{N} W_n(k) r_{mn}(k - \tau_n^l(k)f_s - \tau_m^l(k)f_s)
\]

\[
= \sum_{m=1}^{M} \sum_{n=1}^{N} W_m(k)W_n(k) r_{mn}(k - \tau_n^l(k)f_s - \tau_m^l(k)f_s), \quad l = 1, 2, 3, \ldots, L
\]  

(2)

\(^1\)Another problem can be motion artifacts, but this is not considered here.
where \( W_m(k) \) is the apodization value for transmit element \( m \), and \( \tau'_m(k) \) is the dynamic focusing times for transmit element \( m \) and line \( l \). Since motion compensation is assumed unnecessary, (2) suggests that the high resolution lines can be composed sequentially as the individual transmissions are acquired.

### 3 TEMPORALLY ENCODED MULTI-ELEMENT STA IMAGING

The objective of this study is to investigate a new approach to increase the SNR of conventional STA imaging. It is called *temporally Encoded Multi-Element STA imaging* (EMESTA), and it is based on the combination of multi-element STA imaging with coded excitation waveforms. These two approaches are described briefly in turn in the following.

#### 3.1 Multi-Element STA Imaging

The major drawback of the STA imaging approach is the SNR. Since a single element is used at each transmission the SNR is very low compared to e.g. linear array imaging, which significantly limits its clinical application. To overcome this, a subaperture consisting of multiple grouped elements can be used to emulate the radiation pattern of a single (virtual) element, and hereby increase the SNR. This concept is called *Multi-Element Synthetic Transmit Aperture imaging* (MESTA), and has previously been investigated by Karaman et al. with respect to application in hand-held scanners\[4\], and by Lockwood et al. with the purpose of 3-D ultrasound imaging\[5\]. Both groups show that by properly delaying the individual elements in the subaperture, good approximation to the single element beam can be obtained along with a significant improvement in SNR.

The defocusing delay for each subaperture element can be calculated as illustrated in Fig. 1. A virtual element position or defocusing point is selected with a certain axial distance to the center of the subaperture, e.g. as a fraction of the subaperture size. Given the element coordinates and the center of the subaperture, the defocusing delay \( \tau_d \) for a specific element is then calculated by

\[
\tau_d = \frac{|\vec{R}| - |\vec{r}|}{c},
\]

where \( \vec{R} \) is the vector from the defocusing point to the element, \( \vec{r} \) is the vector from the defocusing point to the subaperture center, and \( c \) is the sound speed. The lengths of the two vectors are calculated from Fig. 1 as

\[
|\vec{R}| = \sqrt{(x_e - x_d)^2 + (y_e - y_d)^2 + (z_e - z_d)^2},
\]

\[
|\vec{r}| = \sqrt{(x_c - x_d)^2 + (y_c - y_d)^2 + (z_c - z_d)^2}.
\]

The position of the defocusing point, which produces the best emulation of the single element radiation pattern, is not straightforward to calculate, because it depends on the interference between the waves from the individual elements.
and their radiation patterns. On one hand, if the defocusing point is moved away from the subaperture, the beam will become increasingly flat and narrow, and at some point yield a plane wave approximation. But, the defocusing point cannot be moved too close to the subaperture either, because the difference between the transmit delays becomes too large to obtain a coherent wavefront. Therefore, the location of a proper defocusing point should be investigated through simulations. Different apodization schemes can be applied on the subaperture to attenuate certain elements and obtain a better approximation[7].

3.2 Temporal Encoding using Linear FM Signals

Linear frequency modulated (FM) signals have been used in radars for decades because of their distinctive feature of serving as both a long and a short pulse simultaneously. In medical ultrasound imaging, the first framework for using coded signals to increase the penetration depth in phased array imaging was described by O’Donnell[1] in 1992. The conventional linear FM signal is given by [8]

$$s(t) = b \cos \left( 2\pi \left[ f_c t + \frac{B_s}{2\tau_s} t^2 \right] \right), \quad |t| \leq \frac{\tau_s}{2}, \quad (6)$$

where $b$ is the amplitude, $f_c$ is the center frequency, $B_s$ is the −6 dB bandwidth, and $\tau_s$ is the duration of the signal. The matched filter to $s(t)$ in (6) is given by

$$h_m(t) = s(-t) = b \cos \left( 2\pi \left[ f_c t - \frac{B_s}{2\tau_s} t^2 \right] \right), \quad |t| \leq \frac{\tau_s}{2}. \quad (7)$$

Assuming $s(t)$ is the waveform received from a point target, the compressed rf signal becomes[8]

$$r(t) = s(t) * h_m(t) = \frac{b^2 \tau_s}{2} \sin \left( \frac{\pi D \tau_s}{\tau_s^2} \left( 1 - \frac{|t|}{\tau_s} \right) \right) \cos \left( 2\pi f_c t \right), \quad |t| \leq \tau_s \quad (8)$$

where $D$ is the time-bandwidth product. Considering only the envelope of $r(t)$, this has approximately the shape of a sinc-function. Assuming $D$ is large (generally above 20 as mentioned by Blinchikoff and Zverev[8]), the envelope of $r(t)$, $r_e(t)$, becomes a sinc-function

$$r_e(t) = |r(t)| = \frac{b^2 \tau_s}{2} \frac{\sin \left( \frac{\pi D \tau_s}{\tau_s^2} \right)}{\pi D \tau_s}, \quad |t| \leq \tau_s, \quad D > 20. \quad (9)$$

The $\frac{\pi D}{\tau_s} \approx -4$ dB width of this function is $\delta t = \frac{1}{D}$, which shows that the temporal (axial) resolution can be improved by increasing the bandwidth. This is advantageous when applying high bandwidth transducers. An example of the compression result in (9) is shown as the dashed curve in Fig. 2 (bottom). The influence of a typical 8.5 MHz linear array transducer has been introduced, and the parameters for $s(t)$ and $h_m(t)$ are $f_c = 7$ MHz, $B_s \simeq 7$ MHz, and $\tau_s = 20 \mu$s. As seen the temporal sidelobes are around -40 dB, which is not sufficient for medical imaging. To reduce the sidelobes, previous research has shown that this can effectively be done by applying amplitude weighting on the linear FM signal and compression filter[2, 9]. In this study, a Tukey window with a 10% duration is applied on the linear FM signal, and a Chebychev window with 70 dB relative sidelobe attenuation is applied on the compression filter ($h_m(t)$). The modified linear FM signal and compression filter are shown in the left and right parts of Fig. 2, respectively. The same parameters as given above have been used. The compression output is shown as the solid curve in bottom figure. As seen the temporal sidelobes have been reduced below -60 dB, which is adequate for clinical imaging.

3.3 Signal-to-Noise-Ratio

The (peak) SNR after matched filtering is directly proportional to the energy in the received signal. Also, the peak compression output in (8) is proportional to the energy in the linear FM signal. Thus, the analysis can be simplified by
only considering the energy, when comparing the SNR obtained by linear array imaging and EMESTA imaging. Also, the noise in the system is assumed white and uncorrelated, and the received signals from each channel are perfectly phase aligned. In this case there is no difference between transmitting with the same aperture $M$ times or with $M$ different apertures with the same size. Thus, the beamformer can be regarded as simply averaging the received channel signals. Furthermore, attenuation and diffraction effects are not included in this analysis, and it is assumed that no amplitude weighting is applied to the transmitted waveforms and receive filters.

For linear array imaging the energy in the received signal for a single element $E_{RL}$ from a point target at the acoustic focus is proportional to the number of transmit element $N_T$ squared and the duration of the transmitted pulse $\tau_t$

$$E_{RL} \sim N_T^2 \tau_t.$$  \hfill (10)

As showed experimentally by Karaman and coworkers [4] the amplitude of the wavefront created by the defocused subaperture consisting of $A_T$ elements is proportional to $\sqrt{A_T}$. Thus, the energy in the received signal for a single element $E_{RS}$ for EMESTA imaging is

$$E_{RS} \sim A_T \tau_s,$$  \hfill (11)

where $\tau_s$ is the duration of the transmitted FM signal. As mentioned above the beamformer is considered as simply averaging the individual channel signals. Let the number of receive channel be $N_R$, then the SNR in the linear array image is proportional to

$$\text{SNR}_{RL} \sim N_R N_T^2 \tau_t,$$  \hfill (12)
since the noise is white and uncorrelated between channels. From the description of STA beamforming, (2) states that in EMESTA imaging averaging is done over $N_R$ receive channels and $M$ transmit events. Thus, the SNR in the EMESTA image is proportional to

$$\text{SNR}_{Bl} \sim MN_R \Lambda T \tau_s.$$  

(13)

Taking the ratio between (13) and (12) yields

$$I_B = \frac{\text{SNR}_{Bl}}{\text{SNR}_{Blc}} \sim \frac{M \Lambda T \tau_s}{N^2 \tau_l}.$$  

(14)

This assumes that the noise is stationary and described by the same probability density function for both systems. If a better SNR is to be obtained, this ratio needs to be greater than 1, thus

$$\tau_s \geq N^2 \tau_l.$$  

(15)

In this study, a subaperture of $A_T = 33$ elements will be used. The applied FM signal has a duration of $\tau_s = 20 \mu s$, and the transducer has 128 transmit and receive elements, and thus the number of transmissions will be $M = 96$. Using (14) and a linear array imaging setup of $N_T = 64$, $N_R = 128$, and a 2 cycle sinusoid at 7 MHz, it is found that $I_B \approx 54.2 \approx 17$ dB. This indicates that a significant increase in SNR is to be expected.

4 SIMULATIONS

The simulations are done using Field II [10] to compare the performance of EMESTA imaging to linear array imaging on a theoretical basis in terms of spatial and contrast resolution.

The transducer model used was a 128 element linear array aperture with a center frequency of 8.5 MHz and a relative bandwidth of 60%. The pitch is 0.208 mm and the element height is 4.5 mm. The transducer has an elevation lens with a focal point at 25 mm. The impulse response of the receive elements is set to the pulse-echo response measured from a plane reflector using a delta excitation. Thus, the impulse response of the transmit aperture is set to a Dirac delta function.

In linear array imaging a 64 element aperture is used for transmission and all 128 elements are used in receive. A 2 cycles sinusoid at 7 MHz weighted with a Hanning window is used as excitation signal, and no transmit apodization is applied. The received signals are filtered using a matched filter, and beamformed using dynamic receive focusing with updated delay curves for every second sample. The number of lines in the image and their spacing is calculated from the highest spatial frequency of the medium to satisfy the spatial Nyquist theorem.

EMESTA imaging is done using a 33 element subaperture and 128 receive elements, which results in 96 emissions. The excitation signal is the linear FM signal displayed in Fig. 2 (top, left). It has a duration of 20 $\mu$s, a center frequency of 7 MHz, and a bandwidth of approximately 7 MHz. The FM signal has been weighted with a Tukey window to reduce the temporal sidelobes. The corresponding compression filter is also shown in Fig. 2 (top, right). A Chebychev window has been applied to reduce the temporal sidelobes in the compression output. The defocusing point is set to $(x_d, y_d, z_d) = (x_c, y_c, d)$, where $d$ is the size of the subaperture, which has been determined through wavefront simulations [7]. Transmit apodization is applied on the subaperture to reduce the influence of edge waves on the point spread function (PSF). Beamforming is done using dynamic transmit and receive focusing. In both linear array imaging and EMESTA imaging, a modified Hamming window with edge levels at -12 dB is applied as receive apodization.

A wire phantom is simulated to investigate the spatial resolution performance. The wires are located at 20 mm to 120 mm with 5 mm spacing. The transmit focal point for linear array imaging was set to 50 mm. Figure 3 shows the -6 dB lateral (top) and axial resolutions (bottom) as a function of depth for linear array imaging (dashed) and EMESTA imaging (solid) respectively. As expected the lateral performance of EMESTA imaging is better after the linear array transmit focus, and it improves with depth due to the application of dynamic transmit focusing. At 120mm the lateral resolution has improved by approximately 30%. The performance of the axial resolutions is seen to be close, but the compression mechanism of the linear FM signal is more stable with depth.

The contrast resolution is evaluated by simulating a cyst phantom containing four cysts at depths of 30, 40, 50, and 60 mm
Figure 3: -6dB lateral (top) and axial (bottom) resolution for linear array imaging (dashed) and EMESTA imaging (solid) respectively. The results are obtained from a simulated wire phantom with 5mm spacing between the wires. The transmit focal length for linear array imaging was set to 50mm.

Figure 4: Simulated cyst phantom for linear array imaging (left) and EMESTA imaging (right) containing four cysts at depth 30, 40, 50, and 60 mm with amplitudes -40, 4, 6, and 12 dB above the surrounding speckle respectively. No attenuation effects are included.

with amplitudes of -40, 4, 6, and 12 dB above the surrounding speckle, respectively. No attenuation effects are included. The results are shown in Fig. 4 for linear array imaging (left) and EMESTA imaging (right). The linear array transmit focus was again set to 50mm.

As a measure of the contrast resolution the contrast-to-noise ratio (CNR) is used as proposed by Ucar and Karaman [11]

$$\text{CNR} = \frac{\mu_c - \mu_s}{\sqrt{0.5(\sigma_c^2 + \sigma_s^2)}}$$

(16)

where $\mu_c$ and $\sigma_c^2$ are the mean and variance of the intensity inside the target, and $\mu_s$ and $\sigma_s^2$ are the mean and variance of the intensity within the speckle at the same depth as the target. Figure 4 shows no conspicuous difference, and this observation is also supported by the calculated CNRs, which are summarized in Table 1. From Fig. 4, no significant difference in the speckle size is observed, which is also in accordance with the results in Fig. 3.

Based on the results presented above, it is concluded that the EMESTA imaging and linear array imaging have equal contrast performance, and EMESTA imaging obtains better lateral resolution after the linear array focal point.

5 MEASUREMENTS

The measurements are performed using the experimental multi-channel ultrasound scanning system, the RASMUS system, developed at our center. The system has 128 individually programmable transmitters capable of sending arbitrary coded

<table>
<thead>
<tr>
<th>Cyst</th>
<th>Linear Array Imaging</th>
<th>EMESTA Imaging</th>
</tr>
</thead>
<tbody>
<tr>
<td>-40dB</td>
<td>1.9241</td>
<td>1.8151</td>
</tr>
<tr>
<td>4dB</td>
<td>0.7420</td>
<td>0.7430</td>
</tr>
<tr>
<td>6dB</td>
<td>1.0950</td>
<td>1.1305</td>
</tr>
<tr>
<td>12dB</td>
<td>1.9801</td>
<td>1.7992</td>
</tr>
</tbody>
</table>

Table 1: Calculated CNR for simulated cyst phantoms.
waveforms with a precision of 12 bits at 40 MHz. Sixty-four receive channels can be simultaneously sampled at 12 bits and 40 MHz, and the 2 to 1 multiplexing in the system enables acquisition of 128 channels in real-time over two transmissions. The RASMUS system is remotely accessible and programmed through a developed Matlab interface. For the measurements a 8.5 MHz linear array transducer is used with 128 elements. The relative bandwidth is approximately 60%, and the pitch is 0.208 mm. The transducer elements have a height of 4.5 mm and an elevation lens with a geometric focal point at 25 mm.

The measurement setups for linear array imaging and EMESTA imaging are the same as those used for the simulations. This has been chosen to enable direct comparison between the measurements and the simulations. The same transmit voltage is used for both linear array imaging and EMESTA imaging, and the filters used for matched filtering and compression of the received signals have been normalized equally to enable a fair comparison. The linear FM signal used in both the simulations and measurements has been designed to obtain temporal sidelobe levels at approximately -60dB. To evaluate this, a wire phantom containing four wires and water has been scanned. Figures 5 and 6 show the B-mode images and axial projections for linear array imaging and EMESTA imaging respectively. The dynamic range in the B-mode images is 60 dB. As seen the temporal sidelobes have been reduced to approximately -55dB, which is adequate for clinical imaging.

The spatial resolution and SNR performance of EMESTA imaging is evaluated in the presence of attenuation using a multi-target phantom with 0.5 dB/[cm MHz] attenuation. Figure 7 shows the linear array image (left) and EMESTA image (right) of a scanned region containing twisted nylon wires spaced axially by 1 cm. The dynamic range is 50 dB in both images. As seen the penetration depth has significantly been increased using EMESTA imaging. The linear array image has a very low SNR after 70 mm, and the wires are not visible in this region. No apparent noise is present in the EMESTA image, and the wires are thus fully visible throughout the imaged region. This indicates an increase in penetration depth of more than 3 cm.

The lateral and axial resolutions are evaluated for each wire in both images in Fig. 7. The results are shown in Fig. 8, where the lateral resolution is displayed left and the axial resolution in the right figure. As seen EMESTA imaging has a better lateral performance at distances after the linear array transmit focus due to dynamic transmit focusing. Compared to the simulation results in Fig. 3, it is noted that the lateral resolutions follow the same trends, but it is a bit higher in...
Figure 7: Linear array image (left) and EMESTA image (right) of a scanned multi-target phantom with 0.5 dB/cm MHz attenuation. The scanned section contains twisted nylon wires spaced axially by 1 cm throughout the imaged region. The dynamic range is 50 dB.

The measurements due to the presence of attenuation. The axial resolution is better for EMESTA imaging throughout the

Figure 8: Lateral (top) and axial (bottom) resolution curves for linear array imaging (dashed) and EMESTA imaging (solid). The curves have been obtained by calculating the spatial resolutions for each wire in Fig. 7.

Figure 9: Calculated SNR improvement obtained by EMESTA imaging in water (dashed) and in a tissue mimicking phantom (solid).
imaged region. This shows that the linear FM signal has a better axial performance in the presence of attenuation than the conventional short excitation pulse. This observation has been reported previously by Misaridis[12]. The improvement in SNR obtained by EMESTA imaging in water and for a tissue mimicking phantom is shown in Fig. 9. The SNR for water was obtained using the wire phantom data from Figs. 5 and 6. For each image and depth the rms value was calculated resulting in two signal vectors. The noise in the system generated by each imaging method was obtained by scanning a water bath with no reflections. For each imaging method the rms value of the resulting noise images was calculated, resulting in two noise vectors. Taking the ratio between the signal vectors and the noise vectors yields the SNR for each imaging method, and the difference between these SNR curves at the wire locations is the SNR improvement shown as the dashed curve in Fig. 9. The SNR improvement in the tissue mimicking phantom was obtained using the parts of Fig. 7 not containing wires and the generated noise vectors from the water bath. The SNR improvement was evaluated at each depth throughout the image, and the result is shown as the solid curve in Fig. 9. A lowpass filter has been applied on the curve to make it more smooth.

For the water SNR the linear array transmit focus was set to 50 mm, and at this depth the SNR improvement is 8 dB. Compared to the theoretical model proposed earlier, this result does no compare exactly. However, this model did not account for diffraction and amplitude weighting of the FM signal and compression filter. Thus, to obtain a closer correspondence to the measurements, these parameters should be included in the model.

At the linear array focal point in the phantom (40 mm) the improvement in SNR is about 4 dB, but it is significantly larger before and after the focus. At 60 mm an improvement of 12 dB is obtained, and then the ratio starts to decrease approximately linearly until it reaches zero at 95 mm. Note, that the SNR can be improved simply by extending the duration of the linear FM signal. This, however, also extends the dead zone in the beginning of the image, since the transmitters are turned on longer.

To evaluate the contrast resolution of EMESTA imaging, a section of the multi-target phantom containing a +3 dB cyst was scanned. The resulting images are shown in Fig. 10 for linear array imaging (left) and EMESTA imaging (right). The cyst has a 4 mm diameter, and it is located approximately at 65 mm. Because of the reduced penetration depth in the linear array image the cyst is not visible here, but for EMESTA imaging the cyst is slightly visible. This indicates that EMESTA imaging has a slightly better contrast resolution in the presence of attenuation, but this is, however, difficult to determine from this single image and cyst. Therefore, more measurements using a more appropriate phantom are necessary to evaluate the contrast performance.

6 CONCLUSION

In this study, a new approach to increase the SNR of conventional STA imaging has been proposed and investigated. The new imaging approach, EMESTA imaging, utilizes a multiple-elements subaperture to emulate a single transmit element by making a defocused transmission. Furthermore, the conventional short excitation pulses are replaced by linear FM signals with amplitude tapering to reduce the temporal sidelobes resulting from the compression on receive. The approach was compared to linear array imaging, and a theoretical analysis showed that a SNR improvement of 17 dB was possible. Simulations with Field II were done using a 128 linear array with 8.5 MHz center frequency and 60 % bandwidth. The EMESTA approach was setup to use a 33 element subaperture for transmission and a 20 µs modified linear FM signal. In linear array imaging a transmit aperture of 64 element and a 2 cycles sinusoid with Hanning weighting was used. The simulation results showed similar axial resolutions, but the lateral resolution was improved for EMESTA imaging due to the application of dynamic transmit focusing. An analysis of the contrast resolution showed similar performance. Measurements were performed using our experimental multi-channel ultrasound scanning system, RASMUS. The results here showed an improvement in SNR between 4 to 12 dB, and a corresponding increase in penetration depth of more than 3 cm. This was obtained while maintaining a better spatial resolution performance.

In conclusion, the EMESTA imaging approach has shown the capability of significantly improving the SNR, and thus the penetration depth, of conventional STA imaging, exceeding that of linear array imaging. This makes this new approach feasible for clinical imaging and possibly real-time 3D volumetric imaging.
Figure 10: Scanned multi-target phantom containing wires and a +3 dB cyst at 65 mm. The linear array image is shown left and the EMESTA image is shown right. Notice again the improvement in penetration depth.

ACKNOWLEDGMENTS

This work was supported by grant 9700883 and 9700563 from the Danish Science Foundation and by B-K Medical A/S, Gentofte, Denmark.

References


