

OPTIMIZED FILTERS FOR DYNAMIC RF ECHO BLENDING IN MULTIPLE FOCAL ZONE IMAGING

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Abstract – For several practical and technical reasons related to e.g. computational cost, hardware complexity, sensitivity to motion artifacts, frame rate and SNR, most medical ultrasound machines use a fixed focus on transmit and dynamic focusing and filtering on receive. To further improve image resolution, multiple focal zones may be used. The resulting A-lines are superimposed (blended) after envelope detection by means of depth dependent weighting. Due to this incoherent superposition, the spatial resolution of the resulting image can only reach the resolution of the images that were superimposed. To overcome the shortcomings of incoherent superposition, we have developed a technique for coherent, filtered blending of the echoes. In addition to echoes representing different transmit foci, echoes from neighboring beam lines may also be considered.

INTRODUCTION

Beam formers in medical ultrasound systems commonly apply a fixed focus on transmit and dynamic focusing on receive. This approach is reasonable, because every A-line results from a single transmission so that tissue motion is negligible. Optimal image resolution is only achieved close to the depth of the transmit focus. Synthetic aperture focusing techniques (SAFT) provide optimal resolution at all depths. Data resulting from many transmit events, however, is needed to reconstruct the echo amplitude for a given position in the imaging plane. Thus, SAFT is susceptible to motion artifacts. Furthermore, it is computationally costly. A robust alternative to enhance the uniformity of image resolution is to acquire N echoes per beam line applying different transmit focal depths. The standard processing used in multiple focal zone imaging forms an A-line from the N echoes by performing a depth dependent interpolation of the echoes after envelope detection (in-

coherent approach). The resulting image cannot exceed the resolution and signal-to-noise ratio (SNR) of the N A-lines from which it is reconstructed (Note that e.g. averaging two A-lines increases the SNR, but at least one of the A-lines is defocused.).

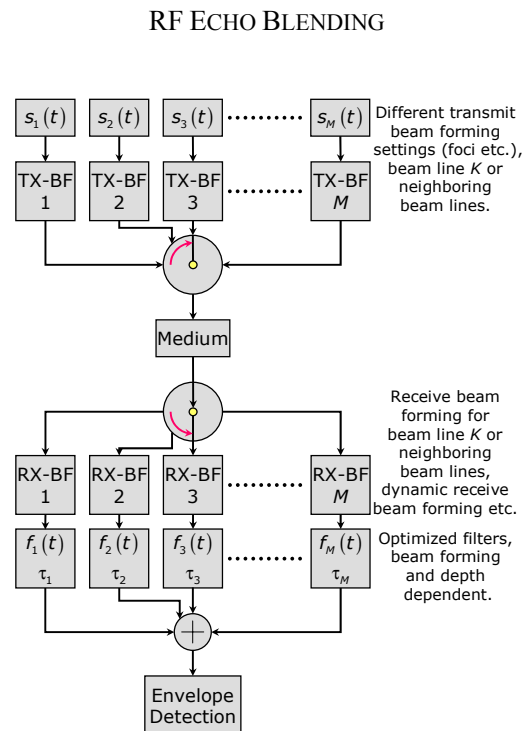


Fig. 1: All M echoes that may be associated with a given beam line K are filtered and then coherently superimposed before envelope detection.

Instead of the incoherent approach described above, we propose to perform a coherent, filtered summation of all echoes $e_k^n(t)$ that may be associated with a given beam line K :

$e_k^n(t)$, where

$$\begin{aligned} n = 1 \dots N : & \quad \text{different transmit foci,} \\ k = K - J \dots K + J : & \quad \text{beam line } K \text{ and} \\ & \quad J \text{ neighboring beam lines} \\ & \quad \text{on either (lateral) side.} \end{aligned} \quad (1)$$

The proposed processing is illustrated in Fig. 1. According to (1), $N(2J+1)$ filters would have to be designed. Beam forming parameters will typically show lateral symmetry with respect to a given beam position. Furthermore, lateral symmetry of the point spread function (PSF) is desirable. Thus, the same filters shall be applied to the echoes

$$e_{K-J}^n \text{ and } e_{K+J}^n, J > 0. \quad (2)$$

The processing proposed in Fig. 1 is linear. Thus, we can add the echoes given by (2) and process the sum with a single filter. The resulting signal is then given by

$$e_k(t) = \sum_{n=1}^N \left[e_k^n(t) * f_1(t) + \sum_{j=1}^J (e_{K-j}^n(t) + e_{K+j}^n(t)) * f_{j+1}(t) \right], \quad (3)$$

where $*$ denotes a convolution.

Filter Optimization

The goal of the coherent, filtered superposition is to optimize the image resolution and/or SNR. The echoes being considered represent different transmit beam forming setting and different viewing angles (neighboring beams). The filters can perform frequency dependent scaling and delaying of the echoes prior to the coherent summation so that an impact on the width of the main lobe as well as on the side lobe, grating lobe and noise level can be foreseen. The filters will have to be depth dependent to account for changes in propagation path length etc.

To find an optimal solution for the filters, the PSF of the system has to be measured or simulated for various depths. For any PSF at a given depth, a core region is defined that represents the desired extension of the main lobe of the PSF. Around this core region, surrounding regions cover those areas, where range (axial), side (lateral) and grating (lateral) lobes are expected, see Fig. 2. The following optimization process determines the filters so that the ratio of signal energy in the core region to signal energy in the surrounding regions is maximal. Different weights may be assigned to the different surrounding regions

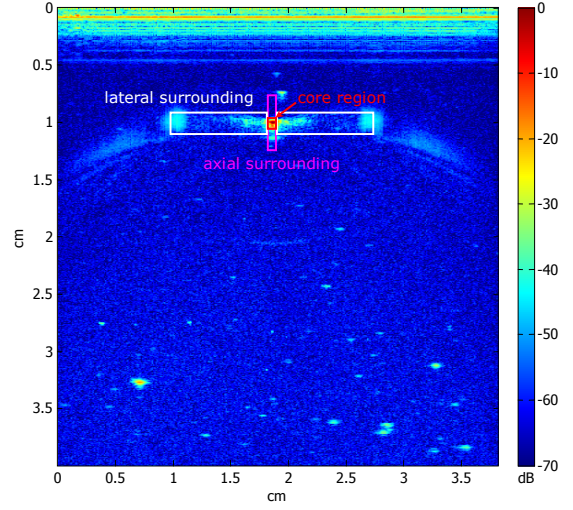


Fig. 2: Placement of regions for optimization of the point spread function (PSF). Measured PSF (absolute value, point approximated by a 10 μm filament).

to e.g. emphasize more on the side lobe than the range lobe reduction. A typical placement of the regions can be seen in Fig. 2.

The actual algorithm for the calculation of FIR filter coefficients was developed by our group for nonlinear or contrast agent imaging, where different transmit pulses instead of different beam forming settings result in a set of echoes representing the same beam line. A description of the algorithm can be found in [1] and [2].

DATA ACQUISITION

Data was acquired using a Siemens Sonoline[®] Antares System equipped with a VF10-5 transducer (linear array, width: 4 cm, center frequency: 7.5 MHz). The system was set up to use 4 foci at 1 cm, 1.5 cm, 2 cm, and 3 cm. 360 beam lines covered a lateral width of about 3.8 cm, i.e. the line spacing is approximately $\lambda/2$. PSFs were measured in a water tank between 1 mm and 37 mm depth in 1 mm steps. The step size was shown to provide smooth transitions between filter responses for subsequent depth. RF data was acquired with 16 bits resolution at 40 MHz sampling rate using the Axis Direct Ultrasound Research Interface.

RESULTS

The following results were obtained with the core region and the surrounding regions placed as shown

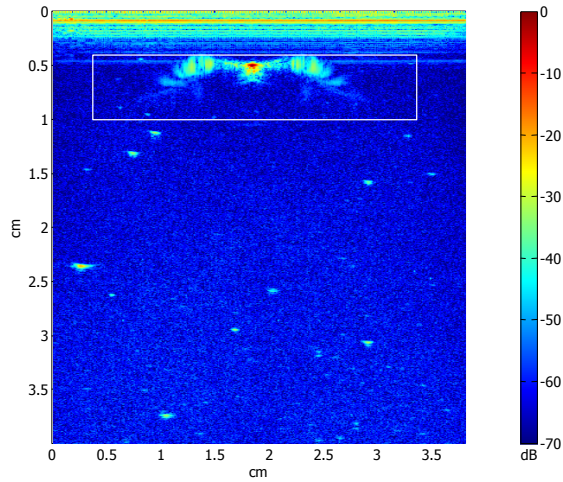


Fig. 3: PSF measure at 0.5 cm. Note the side and grating lobes as well as the asymmetrical representation of the string target in axial direction. The white box shows the region from which beam profiles are derived.

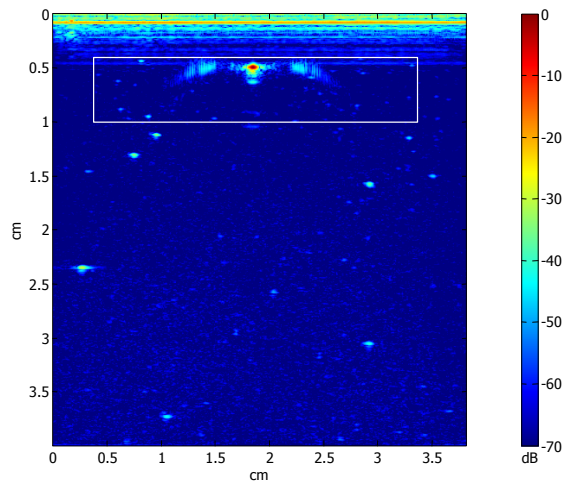


Fig. 4: The proposed processing reduces the grating and side lobes levels significantly and lowers the noise floor by about 4 dB.

in Fig. 2. A practical size for the core regions was found to be $1\lambda \dots 2\lambda$ in both, axial and lateral direction. The filters are FIR filters. A reasonable length turned out to be 24 taps at a sampling rate of 40 MHz. Considering all 4 foci drastically improved the artifact suppression (side lobes and grating lobes) in the near field (1 mm - 20 mm). Grating lobes are less pronounced in the depths beyond 2 cm. Consideration of neighboring beam lines improved the SNR and to some extent the width of the main lobe. This effect can easily be explained by the fact that there is a sig-

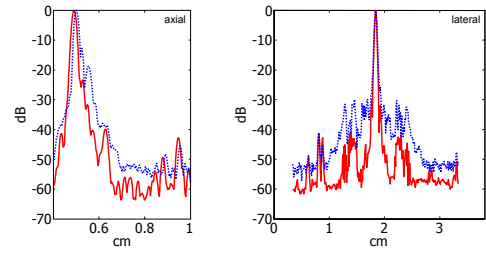


Fig. 5: Beam profiles corresponding to Fig. 3 and Fig. 4 based on a maximum intensity projection within the marked regions (Fig. 3, Fig. 4). Blue, dotted: unprocessed. Red, solid: processed. Note that the processing symmetrizes the PSF in axial direction. Peaks between 0.8 cm and 1 cm are caused by air bubbles. In lateral direction, the grating lobe and side lobe levels are significantly reduced. The beam width decreases slightly. Peaks at e.g. 0.8 cm and 2.8 cm are caused by air bubbles. The noise floor is lower by about 4 dB.

nificant overlap of neighboring beam lines in these depths. $J=1$ (nearest neighbors) was found to be optimal with respect to resolution enhancement vs. computational cost.

In vitro results

Figures Fig. 3, Fig. 4, Fig. 6, and Fig. 7 visualize the improvement of the PSF at 0.5 cm and 2 cm depth. White boxes mark the regions from which the beam profiles were generated (Fig. 5, Fig. 8). Note that the dynamic range of 70 dB is higher than in standard B-mode images, where the artifacts could hardly be seen.

A more quantitative representation is given by the beam profiles, see Fig. 5 and Fig. 8. The beam profiles are maximum intensity projections in lateral and axial direction, respectively, of echo amplitudes taken from the marked regions in Fig. 3, Fig. 4, Fig. 6, and Fig. 7. Note that some peaks in these plots result from air bubbles floating in the water.

In vivo results

Fig. 9 shows an image of the carotid artery and the left lobe of the thyroid. Due to the reduction of grating and side lobe levels, the processed image looks less “hazy”. Vessels appear darker and the overall contrast is improved.

CONCLUSIONS

The proposed technique for RF echo blending reduces artifacts and improves resolution and SNR significantly. The technique can practically be imple-

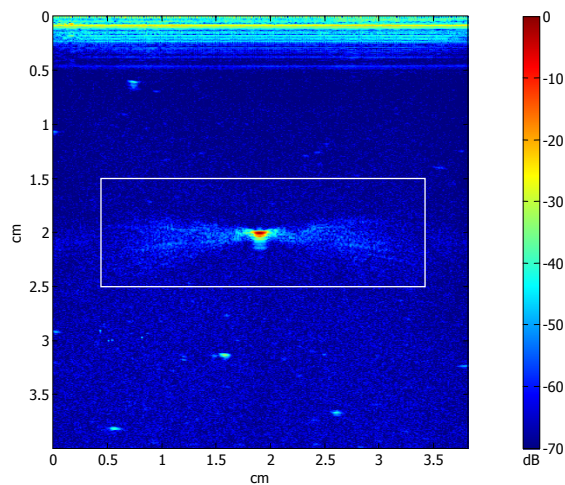


Fig. 6: PSF measure at 2 cm. Note the side and grating lobes as well as the asymmetrical representation of the string target in axial direction. The white box shows the region from which beam profiles are derived.

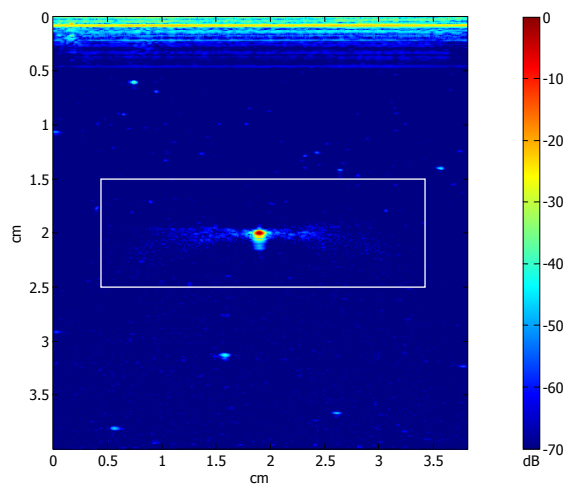


Fig. 7: The processing especially reduces the side lobe level and improves the SNR.

mented in ultrasound systems, since all filters can be stored as presets. The concept also provides a solution for monostatic SAFT based on focused echo data [3, 4].

ACKNOWLEDGEMENT

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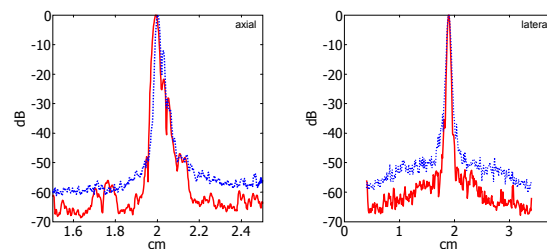


Fig. 8: Beam profiles corresponding to Fig. 6 and Fig. 7 based on a maximum intensity projection within the marked regions (Fig. 6, Fig. 7). Blue, dotted: unprocessed. Red, solid: processed. Note that the processing symmetrizes the PSF in axial direction. Peaks between e.g. 1.7 cm and 1.8 cm in the axial profiles are caused by air bubbles. The processing reduces the lobe levels (lateral). The noise floor is lowered by about 4 dB. Lateral resolution is gained, while axial resolution is lost. Both effects are minor.

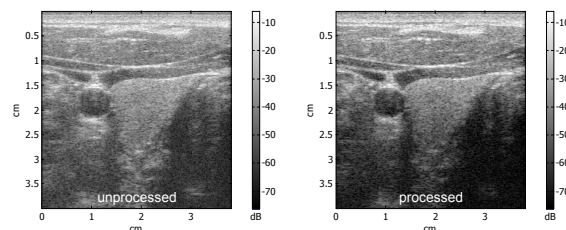


Fig. 9: In vivo images showing the carotid artery and the left lobe of the thyroid. The processing reduces the “haze” so that image contrast is improved.

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