Synthetic Aperture Sequential Beamforming using Spatial Matched Filtering

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Synthetic Aperture Sequential Beamforming using Spatial Matched Filtering

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Abstract—Synthetic Aperture Sequential Beamforming (SASB) has shown to achieve a good resolution and high penetration depth. The low complexity at the transducer level of the beamformer makes it ideal for use with a handheld device. SASB with a low $F\# (\leq 0.5)$ can achieve even better resolution at the cost of high grating lobes, which causes loss of contrast in the final image. In this paper, Spatial Matched Filtering (SMF) was used instead the second stage of beamformer, in an attempt to suppress the grating lobes. The advantage of SMF over SASB was investigated by pushing the limits of $F\#$, from 1.5 to 0.5. The effect of the number of emissions used in first stage was also investigated. A 3.3 MHz BK Ultrasound 9040 convex array was simulated in Field II on a point scatter phantom and a cyst phantom. The resolution was quantified for the cyst mimicking phantom. The results showed that SMF achieved similar resolution as SASB and improved grating lobe suppression leading to an increase in contrast. The grating lobes caused by an $F\#$ of 0.5 are dominant in the SASB images, but not as much in SMF images. The CNR for a cyst mimicking phantom was improved 7 dB and 6 dB for SMF over SASB at depth 20 mm and 30 mm, with an $F\#$ of 0.5 and 256 emissions. The FWHM for SMF was slightly higher than SASB across all depth and parameter settings, with a maximum difference of 0.3 mm. It was demonstrated that SMF can achieve similar resolution to SASB and for certain parameter settings improve the contrast by suppressing the grating lobe artifacts.

I. INTRODUCTION

Synthetic Aperture Sequential Beamforming (SASB) has shown great potential for use in a handheld device setup [1]. In some cases it even outperforms the conventional Dynamic Receive Focusing (DRF) [1]. SASB is a two stage beamforming approach. The first stage is handled at the transducer level with a fixed received beamformer, where the time delay profile and apodization weights are fixed. The focusing scheme is thereby simplified compared to DRF, where the receive delay profile changes dynamically as a function of depth. The output from the first stage beamformer is a low resolution image (LRI) for emission $n$. Adding the next LRI for $n+1$ reveals a HRI that is updated sequentially, as the next LRI is passed through the second stage beamformer. The FRF scheme result in a limited area being hit by the emitted wave. This can be seen in Fig. 1a. An image point at $r_{\theta_n}$ vs $r_n$ for $n=1:2D$ is therefore represented in emission 1 and 2, but will not be present in 3. By changing $r_{\theta_n}$ and $r_{\theta_n}$ between emissions generates the synthetic aperture at positions $r_{\theta_{1:N}}$ for $n=1:N$, where $N$ is the total number of emissions used in the first stage to cover a full scan sector. This has previously been described for a convex transducer array [1].

A. Synthetic Aperture Sequential Beamforming (SASB)

SASB synthetizes a virtual aperture by applying a Fixed Receive Focusing (FRF) scheme as a first stage beamformer. The output from the first stage beamformer is a low resolution line (LRL). The transmission origin ($r_{\theta_n}$) and the fixed focal point, i.e. virtual sources ($r_{\theta_{1:N}}$) are then used to generate an image based on the wave path of the emitted wave [2]. The sequential aspect of SASB is achieved by generating a single 2D low resolution image (LRI) for emission $n$. Adding the next LRI for $n+1$ reveals a HRI that is updated sequentially, as the next LRL is passed through the second stage beamformer. The FRF scheme result in a limited area being hit by the emitted wave. This can be seen in Fig. 1a. An image point at $r_{\theta}$ is therefore represented in emission 1 and 2, but will not be present in 3. By changing $r_{\theta_n}$ and $r_{\theta_n}$ between emissions generates the synthetic aperture at positions $r_{\theta_{1:N}}$ for $n=1:N$, where $N$ is the total number of emissions used in the first stage to cover a full scan sector. This has previously been described for a convex transducer array [1].
B. Spatial Matched Filtering (SMF)

The second stage of SASB can be substituted with an SMF approach. The first stage consist of a FRF beamformer like the conventional SASB \[1, 2\]. Similarly, only the insonified area seen in Fig. 1a is considered as targets for filtration. For each point inside the insonified area, the response of each FRF beamformer are matched filtered using a time-reversed version of the expected response. Finally the SMF responses of \(N\) emissions are summed to reveal the final HRI. The final SMF HRI can be sequentially updated, by repeating the emissions scheme. A graphical representation of the algorithm is found in Fig. 1b. The method assumes the linear ultrasound model is valid and that the optimal filter is given by the time-reversed expected response from a ”true” scatter point placed at \(\vec{r}_i\) and also accounts for the effect of the transducer array geometry including their focusing and apodization.

The output from the first stage beamformer, i.e. \(LRL\) acquired from emission \(n\), is denoted noted \(LRL_n\). \(\vec{r}_{\theta_n}\) and \(\vec{r}_{\epsilon_n}\) can be changed with a sliding aperture approach. The time delay and apodization weighs used to beamform \(LRL_n\) should therefore match the exact same as \(E_{FRF,n}\). \(t_i\) and \(\Delta T_i\) are provided, when \(E_{FRF,n}\) are estimated. \(E_{FRF,n}\) can be estimated with a simulation or experimentally recorded. For each emission \(n\) a total of \(M\) simulations are required, where \(M\) is the number of image points in the defined image grid.

II. Method

The model used in this paper was the BK9040 convex transducer array. The method is easily translatable to other transducer geometries. The developed algorithm was tested with simulations performed with Field II \[5, 6\]. The phantoms used consisted of a point scatter phantom and a cyst-mimicking phantom. To preserve the portable aspect of SASB, the second stage of SASB was substituted with an SMF approach. SMF was applied on the \(LRLs\), to ensure the low complexity of the first stage beamformer. This ensures only one signal is passed from the first to the second stage per emission, similar to SASB. The enhanced resolution with a lower \(F\#\) for SASB was the foundation of exploring the feasibility of applying SMF. To utilize the gain in resolution, the side and/or grating lobes should be decreased or completely removed.

A. Simulation Setup and Image Performance estimators

\(E_{FRF,n}\) for the SMF algorithm were simulated with Field II, providing the match filter coefficients. The simulations used to test the algorithm were likewise simulated with Field II. The
Fig. 2. Left panel: shows contour plot of PSF at 60 mm imaged with SASB, for different F#. Right panel: contour of PSF at 60 mm imaged with SMF, for different F#. All were acquired with 384 emissions. The contour lines are shown with a spacing of 12 dB, and the side lobe and grating lobe peak values are reported for F# = 0.5.

The current form of the SMF algorithm revealed higher sidelobes, as seen by the mirrored side lobes in Fig. 2. Here contour plot of the Point Spread function at 60 mm is seen. Grating lobes were present for an F# of 0.5 for SASB and suppressed with SMF. The M2Gp measure revealed a higher grating lobe suppression for N = 384 and F# of 0.5, as seen in Fig 3a. The quantified lateral FWHMx and 20 dB cystic resolution (CR20db) are shown in Fig. 3b. These revealed that both SASB and SMF follows the same pattern, of increased resolution with decreasing F#. The largest difference between the two were 0.3 mm and occurred at 30 mm for F# of 1.5, i.e. the focal point of the FRF in the first stage. Both SASB and SMF outperforms DRF in FWHMx for F# ≤ 1. The CR20db for F# = 0.5 is comparable to DRF. Due to the hamming apodization, the side lobes are quite dominant when the F# is lower than 1 resulting in a higher CR20db. The mirrored side lobes of SMF provided the generally higher CR20db for SMF compared to SASB. The cyst phantom was imaged with SASB and SMF in Fig. 4. The CNR was quantified for the cyst at 20 and 30 mm depth. For SMF, a CNR of 8.21 dB, and 2.70 dB, compared with 1.12 dB, and −3.90 dB provided by SASB. This gives a CNR improvement of approximately 7 dB at 20 mm, and 6 dB at 30 mm. The processing time on a Intel Core i7-3770 CPU @ 3.40GHz x 8 CPU, with 7.8 Gb ram provided an increase in processing time of 45 minutes of SMF in Matlab, compared to SASB for the images in Fig 4.
Fig. 3. a): top shows the measured M2Gp for N = 192, 256, 384 with $F\# = 0.5$, and bottom shows how the measure is extracted from a PSF profile at radius $R$ from the transducer surface. b): Shows the calculated FWHM, and CR20dB for SMF and SASB with $F\# = 1.5, 1, 0.5$. SMF is shown with red, SASB with black, and DRF is shown with blue color.

Fig. 4. Both images were made of the same Field II simulated data originating from a cyst mimicking phantom. Top image shows SMF, bottom image shows SASB. Dynamic range is 60 dB. Images were acquired with 256 emissions, $F\# = 0.5$ and a scan sector of 60°.

IV. CONCLUSION

In this paper we demonstrate that an SMF approach can be successfully used in place of the second stage beamformer of a synthetic aperture approach with simulations. The resolution proved to be similar to standard SASB for the given parameters. An improved contrast can be achieved with SMF over SASB for an $F\#$ of 0.5 in simulations. The area of application depends on the purpose, and the higher processing time has to be considered, when deciding between SASB and SMF. Simulation and the correlation computation time should be considered as obstacles for a commercial implementation of the method.

REFERENCES