# APERTURE DOMAIN MODEL IMAGE RECONSTRUCTION (ADMIRE) FOR IMPROVED ULTRASOUND IMAGING

## B. Byram

Vanderbilt University Biomedical Engineering Nashville, TN, USA email: brett.c.byram@vanderbilt.edu

## ABSTRACT

In some patients, clinical ultrasound produces beautiful images, while in other patients the images are unusable. This wide variability in outcomes exists for all standard beamformers ranging from delay-and-sum to adaptive methods like the MVDR beamformer. We have demonstrated that a model based approach called Aperture Domain Model Image REconstruction (ADMIRE) consistently improves image quality. ADMIRE works by explicitly accounting for various types of clutter. The algorithm preserves the quality of the best images and produces 10-20 dB improvements in low-quality images.

Here we present results related to beamforming in the presence of multipath scattering—or reverberation–clutter. In the presence of this type of clutter canonical apodization methods fail to improve image quality, but by applying AD-MIRE first the expected improvements from apodization can be restored.

*Index Terms*— Ultrasound, Beamforming, Clutter, Reverberation, Model

## 1. INTRODUCTION

The medical applications of ultrasonic imaging produces a huge range in quality. Manufacturers regularly present amazing images acquired from so-called "glass-walled" subjects. Occasionally, this is reproduced in patients; however, most clinical images are less pristine, and in a fair number of cases ultrasound image quality is so poor that the results are non-diagnostic. Bad images can be caused by a number of mechanisms. The mechanisms resulting in any specific non-diagnostic scenario are typically unclear, but many different mechanisms of degradation have been implicated in general. These include attenuation, diffraction limitations, bright off-axis scattering, sound-speed deviation, and multipath scattering [1, 2, 3, 4, 5, 6, 7]. A degrading mechanism of

note is multipath scattering, which has largely been ignored for decades, but it has recently gained renewed interest [7].

With the intent of suppressing multipath clutter, we recently introduced a model-based approach called aperture domain model image reconstruction (ADMIRE) [8, 9]. However, the approach is flexible because the choice of model is flexible and may be adapted to address the most likely sources of degradation. In our earliest models we highlighted the reverberation portion of the models. We show two examples of this in Fig. 1. More recently we have created models intended for applications dominated by off-axis clutter.

Here we show that ADMIRE can enhance even simple beamforming techniques like aperture weighting. Specifically, we show that in the presence of multipath scattering receive apodization fails to perform in the canonical manner, but after the application of ADMIRE we achieve the expected improvements.

### 2. METHODS

We start with a very generic signal model that describes the narrowband response to a scatterer from an arbitrary location within the shadow of the aperture. The model is

$$p_s(x;t,\omega) = \sum_{n=0}^{N-1} A(x;x_n,y_n,z_n,\tau_n,\omega) e^{jk\tau(x;x_n,y_n,z_n,\tau_n)},$$
(1)

where k is the wavenumber, x is the aperture position, t and  $\omega$ localize the signal in time and frequency,  $\tau(x; x_n, y_n, z_n, \tau_n)$ is the wavefront delay for a signal arriving from  $(x_n, y_n, z_n)$ at  $\tau_n$ .  $A(x; x_n, z_n, \tau_n, \omega)$  is the lateral amplitude modulation induced by the STFT and the element sensitivity.  $A(x; x_n, z_n, \tau_n, \omega)$  also depends on the signal's pulse shape and  $\tau(x; x_n, y_n, z_n, \tau_n)$ . Here, we express the model generically including the out of plane dimension, y, to indicate that it can be used to account out of plane clutter as well.  $\tau_n$ is a particularly important parameter because it allows us to move away from the standard time-of-flight paradigm typically used in ultrasonic beamforming. That is, typically, one

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(a) In Vivo Example #1

(b) In Vivo Example #2

**Fig. 1:** Matched delay-and-sum and model-based B-mode images from three *in vivo* examples illustrate improved contrast achieved by our model. Data were acquired on a Siemens S2000 (Siemens Healthcare, Ultrasound Business Unit, Mountain View, CA) so the advanced method's limited field of view is indicated by the arrows. All images shown with 70 dB dynamic range.

assumes a linear description of the wavefield, which allows the direct connection between propagation time and depth through the relation  $d = \frac{ct}{2}$ . This relationship no longer holds in the presence of multipath scattering-the source of reverberation clutter.

Based on where we believe clutter originates from we can populate a model matrix, A, with predictors that represent scatterers from clutter generating regions and scatterers from the region of interest. The number of predictors in A is typically large relative to the amount of data so the system is ill-posed. Therefore, we impose a regularization constraint. An additional challenge with reverberation is that the model predictors from different depths are colinear. In order to address this, we used elastic-net regularization. The optimization problem is then,

$$\hat{\beta} = \arg\min_{\beta} (\|y - A\beta\|^2 + \lambda(\alpha \|\beta\|_1 + (1 - \alpha) \|\beta\|_2^2/2)).$$
(2)

We have described the form of y, A, and  $\beta$  elsewhere, along with the specific advantages provided by joint L1, L2 regularization [9].

We specified a generic form of a narrowband scattering model in (1). To apply our approach to broadband ultrasound data without losing axial resolution, we use the short-time Fourier transform (STFT) and apply the algorithm to the individual frequency bands within the bandwidth of the pulse. After the decluttering the data, we can use an inverse STFT (ISTFT) to create an estimate of the time-domain signal corresponding to the STFT data.

#### 2.1. Simulations

We used Field II simulations to quantify the connection between apodization and reverberation clutter [10]. We generated channel data of a 4 mm diameter anechoic cyst in a uniform background simulated using a center frequency of 3 MHz, 60% bandwidth and an F/2 imaging system.

In order to generate reverberation clutter, we used a pseudo non-linear approach described previously [11]. The method allows us to use an efficient linear ultrasound simulation tool to quickly generate channel data with the same characteristics as reverberation clutter described in the literature [7] Realizations of reverberation were added to the anechoic cyst channel data. The clutter was added after being scaled relative to the cyst data to create specified signal-toclutter ratios (SCR), calculated as

$$SCR = 20\log_{10} \frac{\sum_{n=0}^{N-1} S_n^2}{\sum_{n=0}^{N-1} C_n^2}.$$
 (3)

 $S_n$  and  $C_n$  are the signal and clutter data, respectively, indexed by channel.

We simulated anechoic cysts with SCR of -20, -10, 0, 10, and 20 dB with four realizations of scatterers for both the cyst and clutter for each case.

### 2.2. Evaluation

We applied ADMIRE to the simulation data and reconstructed channel data. Then, a Hamming apodization function was applied to the pre- and post-ADMIRE receive channel data. This resulted in four beamforming scenarios.



(a) No Apodization

ion (b) Hamming Apodization

(c) ADMIRE–No Apodization

(d) ADMIRE-Hamming

**Fig. 2:** An example of the four different beamforming scenarios are shown for an SCR of 0 dB. The results are shown with 50 dB dynamic range. Qualitatively, the delay and sum cases show little change with or without apodization. The ADMIRE case shows a moderate level of improvement for the dynamic range shown.

The simulation data were evaluating using the contrast and contrast-to-noise ratio metrics applied to the envelope of the RF data before amplitude compression. We calculated the contrast and CNR as

$$C = -20 \log_{10} \left( \frac{\mu_{lesion}}{\mu_{background}} \right), \tag{4}$$

the CNR as

$$CNR = 20\log_{10}\left(\frac{|\mu_{background} - \mu_{lesion}|}{\sqrt{\sigma_{background}^2 + \sigma_{lesion}^2}}\right), \quad (5)$$

where  $\mu$  and  $\sigma^2$  are the mean and variance of regions inside and outside the anechoic cyst.

#### 3. RESULTS

We start by showing one of the simulation data cases with an SCR of 0 dB in Fig. 2. The example shows the cases beamformed with and without Hamming apodization and with and without ADMIRE. The Hamming apodization provides no visual improvement to the delay and sum data, but it does generate modest improvements when applied to the ADMIRE data.

We summarize the outcome of apodization using a series of boxplots for the contrast and CNR metrics in Fig. 3. The boxplots show the change in the respective image metric after applying the hamming apodization. This set of results show several things. First, as expected, the hamming window applied to delay and sum produces about a 4 dB improvement in contrast on the minimally cluttered data. This is of course a canonical result, but it is interesting to note that the effect of apodization actually increases when the window is applied in conjunction with ADMIRE regardless of the clutter level. The mechanism for this substantial additional improvement is unclear, but it is consistent with *in vivo* observations [9]. Second, as the level of clutter increases (i.e. the SCR goes down) the impact of apodization rapidly decreases until apodization does not change the image at all. This appears surprising because the standard expectation is that in the presence of high levels of clutter apodization provides a more substantial improvement in image quality. However, this is only if the clutter originates from bright off-axis structures. In the presence of multipath scattering, apodization is an ineffective strategy for improving contrast. These observations are easiest to see on the contrast result, but they are consistent with the CNR results as well. The CNR results are harder to interpret because we have chosen to display the full range of all of the boxplots, but as a percentage the CNR results are comparable except for the lowest SCR values.

## 4. DISCUSSION AND CONCLUSION

The expectation in clinical ultrasonic imaging is that apodization improves image contrast with a modest trade off in resolution. Our results showed that apodization does not improve contrast and CNR in the presence of sufficient amounts of multipath clutter. Apodization applied in the presence of multipath scattering did not result in worse contrast, which is some consolation, but it is our expectation that the apodization function still reduces resolution. This is something we still need to explore, but given the current evidence it is possible that in difficult to image patients with high levels of clutter apodization may only decrease image quality by reducing resolution.

The primary caveat to the results shown here is that there is little understanding of what levels of multipath clutter are encountered clinically. Our results show that multipath clutter stops resulting in improvements around an SCR of 10 dB. Our anecdotal experience leads us to hypothesize that in many difficult to image patients that the SCR is at least as low as 10 dB, but there is still little quantitative data in the literature to broadly support this assertion.



**Fig. 3:** The results are presented as the improvement obtained by introducing a Hamming apodization window. There is a distinct improvement in contrast for both ADMIRE and delay and sum for high SCR, which is a canonical result. The improvement in CNR is also comparable as a percentage change because the CNR is in the 3-4 dB range relative to contrast in the 20-30 dB range.

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