Design and Development of Tissue Aware Beamformer Architecture for Ultrasound Imaging Applications

Jayaraj U. Kidav, Shankar, and S. G. Sreejeesh

Abstract Ultrasound imaging systems generally use delay and sum (DAS) beamforming scheme for generating the Brightness-mode (B-mode) image from the RF-echoes reflected back from the tissues. However, the images produced from DAS has poor contrast and low resolution. For mitigating the disadvantages offered by DAS, the Delay Multiply and Sum (DMAS) beamforming scheme was introduced. It generates images having superior contrast and improvised resolution. In case of DMAS, information generated from the second harmonics has to be preserved in order to generate the final image. Therefore, the ultrasound imaging system's sampling frequency must accommodate second order echoes which may not be possible in all the cases. Apart from this, the hardware requirement is comparatively more for DMAS scheme because of its computation complexity. However, if US systems could give the flexibility to reconfigure beamforming while taking into account the limitations of DAS and DMAS beamforming schemes, it would be feasible to pick the best beamforming based on system constraints and clinical requirements. A novel algorithm for such tissue aware beamformer is introduced and along with this the architecture for the same is proposed in this paper.

Keywords Delay and sum · Delay multiply and sum · Tissue aware beamforming · Ultrasound imaging · Beamforming architecture

J. U. Kidav
NIELIT Aurangabad, NIELIT Complex, Aurangabad, Maharashtra, India
e-mail: jayaraj@nielit.gov.in

Shankar · S. G. Sreejeesh (✉)
NIELIT Calicut, Calicut, Kerala, India
e-mail: sreejeesh@nielit.gov.in

P. Shivakumara et al. (eds.), *Proceedings of the NIELIT's International Conference on Communication, Electronics and Digital Technologies*, Lecture Notes in Networks and Systems 1522, https://doi.org/10.1007/978-981-96-9932-2_4

1 Introduction

Ultrasound is commonly used to describe mechanical vibrations with oscillation frequencies greater than 20 kHz (i.e., the maximum audible frequency). The phrase ultrasonic imaging refers to the field of distant sensing in which mechanical vibrations with defined characteristics are produced, transmitted across a medium, and successively recorded. In medical applications, ultrasound often refers to longitudinal waves with frequencies ranging from 1 to 50 MHz. Diagnostic ultrasound offers significant benefits over other medical imaging techniques. It uses no ionizing radiation, is non-invasive, and the majority of the exams are painless for the patient. Soft tissues may be seen using ultrasound. Although the visual quality is lower than that of a CT scan, ultrasonic imaging offers the advantage of depicting the dynamics of anatomical structures.

Beamforming is a signal processing method used in the sensor arrays to transmit or receive directed signals [1]. It is also called as spatial filtering. It is accomplished by arranging components in an array of antenna so that signals experience constructive interference at specific angles while others suffer destructive interference. Beamforming may be used to provide spatial selectivity at both the transmitting and receiving ends. The array's directivity is defined as the improvement over omnidirectional reception/transmission.

In the commercial ultrasound imaging systems, the delay and sum (DAS) [2] beamforming method is the most prevalent and widely used. But over the years, many other non-linear and adaptive beamforming schemes have been developed and one such example is Delay Multiple and Sum (DMAS) scheme [3]. Upon thoroughly exploring both the schemes, a novel tissue aware beamformer with the capability to select the optimum beamforming scheme in accordance with the clinical requirements is presented.

2 Background

2.1 Delay and Sum (Das) Beamforming

Before being employed for ultrasound imaging, the delay and sum (DAS) [2] approach was associated with airborne and ground-based radar, along with communications. Its origins may be traced back to steerable array antennas used in shortwave transmission. An antenna array, like a transducer used for medical ultrasound, is described as a set of linked individual antennas that work together.

Algorithm wise it can be represented as follows:

$$\mathrm{YDAS(t)} = \sum_{i=1}^{N} W_{i,k} * s_{i,k}(\mathrm{t}), \forall \mathrm{k} \in \mathrm{S} \tag{1}$$

$$s_{i,k}(t) = x_{i,k}\left(t - \tau_{i,k}\right) \tag{2}$$

where, N is the number of sensor elements or channels, k is the beamformed scan line in the field of view (FOV) or scan area S, $x_{i,k}$ is the ultrasound sensor outputs for i^{th} channel and $s_{i,k}$ is the corresponding delay ($\tau_{i,k}$) compensated signal. Finally, the delay compensated ($s_{i,k}$) signals are multiplied by apodization coefficients ($W_{i,k}$) to form the beamformed output $Y(t)$. This process is repeated until each focal points given in the region of interest are beamformed. Following that, the post processing operation is done, where scan conversion, log compression, etc. can happen and get the final B-mode image.

The DAS has several advantages:

(1) DAS is based on the concepts of fundamental wave propagation (linearity, straight-ray propagation, and weak backscattering);
(2) easy to implement and parallelizable;
(3) Quantitatively sturdy, quick, and real-time applications compatible and
(4) As it is data-independent, the temporal coherence and statistical properties of real envelopes are preserved [2]. However, the problem with DAS technique is that because of higher sidelobe levels, the final image obtained after processing has clutters and significantly less contrast.

2.2 Delay Multiply and Sum (DMAS) Beamforming

If a single scan line is taken, each transducer in the receiving aperture gets a separate echo signal that has travelled along a distinct path after beam transmission. The initial phase of the method is to time-shift the incoming RF signals in order to realign them, like in DAS [3]. As the signals are in phase, they are combinatorically linked and multiplied: assuming the number of receive channels is N, then the number of multiplications to be done is provided by all conceivable signal pair combinations, as shown in Eq. (3).

$$\frac{N}{2} = \frac{N^2 - N}{2} \tag{3}$$

and the beamformed DMAS output is detailed in (4)

$$Y_{DMAS,k}(t) = \sum_{i=1}^{N-1} \sum_{j=i+1}^{N} S_{i,k}(t) S_{k,k}(t) \tag{4}$$

where $S_{i,k}(t)$ is the delay compensated signal of the kth focal point and Y*DMAS*, (t) is the beamformed output. The resulting signal cannot be processed to make the scan line of a standard B-mode picture as it is because owing to the multiplication stage, it is a squared dimensionally (i.e. [Volt]^2, instead of [Volt]), which results into a rectified non-zero mean signal, and in turn making envelope detection not possible. The output of DAS beamforming is a zero-mean signal with the Magnitude spectrum identical to that of the RF signals, but the magnitude spectrum of the DMAS beamformed output has both the second harmonic and DC component. This is because by multiplying two RF signals with almost identical frequency content (e.g., a band centered at), two additional components are formed in the resulting signal's amplitude spectrum one centered at fO and another centred at $fO + fO = 2fO$. Thus, an additional step is included in the DMAS processing chain where the beamformed signals for attenuating the DC and high frequencies components while keeping the ones centered around $2fO$ intact by filtering the output through a band-pass (BP) filter and because of this it is called as Filtered-DMAS (F-DMAS) [3].

The advantages F-DMAS offer while compared with DAS are [3]: the lateral resolution of the picture will improve as the wavelength and f-number (F#) decrease. Because the spectrum of F-DMAS beamformed signal is centred at 2FC, the wavelength is halved in comparison to DAS. Simultaneously, because the auto-correlation function has 2N-1 coefficients, the number of elements in the new "fake" aperture is raised, and F# is reduced. The picture axial resolution, on the other hand, will remain same while the fractional bandwidth is halved as the central frequency is doubled. By introducing an amount of backscattered signal coherence into the beamforming process, an improvement in the noise rejection and clutter is made by the correlation operation. The output of each multiplication operation is amplified when the both the inputs are high, and it is lowered by a significant amount when both the inputs are low, and the contribution from the outlier uncorrelated samples is reduced in a significant way. Furthermore, as compared to simple DAS, many more signals are combined for computing the final output, in turn improving the signal-to-noise ratio (SNR). The contrast resolution (CR) is the beamformer ability to detect targets with different echogenity even during the presence acoustic clutter, such as side/grating lobes; it is usually associated with the pulse-echo beamwidth at −40 dB or lower levels). Because of the multiplication stage (point 2) and the auto-correlation window shaping, in the DMAS the CR is higher.

One of the major issues with the F-DMAS algorithm is its computational complexity $O(N)^2$ which makes it difficult to implement the algorithm on the real-time basis. So, researchers in [4, 5] have developed a technique to bring down the computational complexity to O(N) and O (1) respectively. "Table 1" shows presents the comparison between DAS, F-DMAS and modified DMAS approaches in terms of computational complexity. Here NC is no. of transducer elements.

While the literature has been mainly focused on the creation of novel beamforming algorithms [6–13], a little effort [14] has been done to identify the optimal beamforming scheme or the use of several beamforming schemes in a simultaneous manner for a clinical scenario. Furthermore, the beamforming algorithms developed thus far are solely dependent on imaging information relevant to the individual therapeutic application. Before deployment, it is required to examine the constraints like computing complexity, frequency range of the beamforming techniques, and the

Table 1 Comparison of DAS, DMAS and their modified versions in terms of computations for each focal point given in the region of interest

Algorithm	No. of additions	No. of multiplications
DAS [2]	N_C-1	0
DMAS [3]	$2^*\ N_C-1$	$(N_C^2-N_C)/2$
DS-DMAS [4]	N_C-1	N_C-1
Modified DMAS [5]	$2^*\ N_C-1$	1

possible picture quality. However, as they are opposed goals, a trade-off must be made after considering priorities.

3 Development of Tissue Aware Beamfomring Algorithm

The main parameters which were considered while developing algorithm are the system sampling frequency FS, transducer centre frequency FC, and the maximum permissible penetration depth of the sound wave in the tissue D and its value is typically 300 λ [15] where λ is the wavelength for that particular transducer frequency. Based on the parameters, we proceed to define the beamforming selection scheme parameter P.

$$P = 1; (F_S < 4 * F_C)||(F_S >= 4 * F_C \&\& Z > (D/2)) \tag{5}$$

$$P = 0; (F_S >= 4 * F_C \&\& Z <= (D/2)) \tag{6}$$

If the value of $P = 1$, then the corresponding pixel will be beamformed via DAS and if $P = 0$, then it will be beamformed via DMAS beamforming scheme. Here, Z is the position of the pixel in the z-axis Here we are considering 4*FC instead on the Nyquist Rate of 2*FC because when we beamform the pixel via DMAS, the effective frequency generated is 2*FC. As a result of this, the effective penetration depth D also gets reduced to half of its value i.e. D/2 = 150 λ where λ is the original wavelength of the sound travelling in a tissue from a transducer having a centre frequency of FC. "Figure 1" shows the high-level diagram of the proposed beamforming scheme.

4 Simulation Setup

For validating the proposed beamforming scheme, we used MATLAB®2021.1 and the ultrasound RF datasets required for performing the simulations were obtained from PICMUS [16].

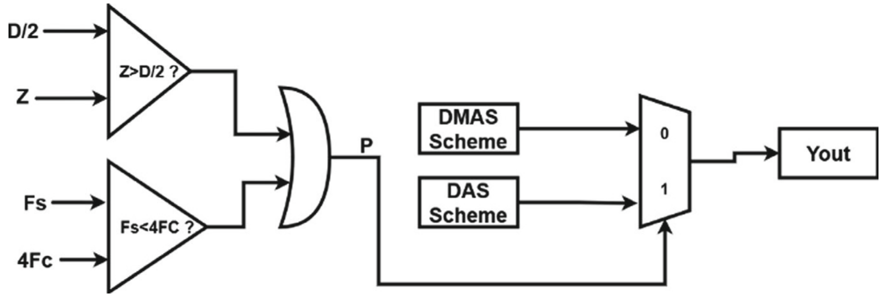

Fig. 1 Proposed beamforming scheme

4.1 MATLAB Simulation Setup

With the acquired RF-signals, we proceeded to the beamforming with the proposed scheme and obtain the final B-mode image. "Figure 2" shows the flowchart of the image B-mode image generation through the proposed pixel-level beamforming algorithm.

Here, we first start of by defining the scan and array parameters as discussed in the section. Once we have loaded the phantom data, we calculate the receive delay for each pixel given in the region of interest (ROI). Then with the help of interpolation, we do the delay compensation of the RF data. After that, we proceed towards the beamforming scheme selection. Once the value of P is generated for the pixel, it gets beamformed via DAS or DMAS based on its value. Once all the pixels beamformed, we proceed for the post processing part, where we do the scan conversion, log compression, etc. to obtain the final B-mode image.

4.2 Results and Analysis

Two RF datasets were considered where the first one is of numerical type used for measuring the contrast resolution (CR) and contrast to noise ratio (CNR). Both the CR and CNR were calculated with the help of the evaluation framework that was available along with the datasets on PICMUS [16]. The other dataset was of in vitro type. In our algorithm, we have previously mentioned that $D = 150\,\lambda$ which effectively translates into a depth of approximately 42 mm. The original dataset image generated from Field II [17, 18], along with DAS, DMAS and Tissue Aware beamformed images for both the datasets are presented here.

In "Fig. 3" we can see the simulation results obtained by performing different beamforming operations on the RF-Data obtained from in vitro type-1 [16]. The original image contains a set of scatter points along with a bright spot which are to be visualized in the final B-mode image. In the DAS scheme, due to high level of sidelobes and clutters, the scatter points and the spot are not distinguishable. In