A combination of pixel-based beamforming and Wiener filter to improve ultrasound image quality

Hui-Wen Xie

*The School of Biological Science
and Medical Engineering
Southeast University
Nanjing, China
huiwen_xie@seu.edu.cn*

Hao Guo

*The School of Biological Science
and Medical Engineering
Southeast University
Nanjing, China
haogseu@seu.edu.cn*

Guang-Quan Zhou

*The School of Biological Science
and Medical Engineering
Southeast University
Nanjing, China
guangquan.zhou@seu.edu.cn*

Nghia Q. Nguyen

*Department of Engineering,
University of Cambridge
Cambridge University - Nanjing Centre
of Technology and Innovation
Cambridge, United Kingdom
nqn20@cam.ac.uk*

Richard W. Prager

*Department of Engineering,
University of Cambridge
Cambridge University - Nanjing Centre
of Technology and Innovation
Cambridge, United Kingdom
rwp@eng.cam.ac.uk*

Abstract—We previously developed coherent pixel-based beamforming to generate two-way focused ultrasound images, where the transmit focusing is achieved by accumulating data from multiple transmits. Most of the impact of this, however, is on the lateral resolution which directly benefits from the large “effective” aperture. In this study we extend the field pattern analysis and form a new Wiener filter that can be combined with coherent pixel-based processing to enhance the axial resolution. The Wiener filter is applied to the received waveform before performing the spatiotemporal mapping. The proposed combination is evaluated through simulation and on data obtained in a phantom study. The imaging results show that the new combination improves on the axial resolution of the coherent pixel-based approach while still maintaining its good performance in lateral resolution and imaging contrast.

Index Terms—ultrasound, pixel-based, Wiener filter, image quality.

I. INTRODUCTION

Advances in high performance computing and hardware flexibility have enabled the implementation of complex ultrasound beamformers as alternatives to conventional dynamic focusing. Prominent among them is the *synthetic aperture* (SA) beamformer [1], which generates an image focused in both transmit and receive (i.e. two-way focusing). The transmit focusing is achieved by accumulating data among multiple transmits acquired using a diverging wave, where the waveshape is usually assumed as a spherical pulse propagating from a single or multiple active elements.

Recently, we have implemented a pixel-based beamformer to enhance the ultrasound image quality [2]. Our method is similar to SA but applied to data acquired through the pulse-echo sequence using a focused transmit beam. Using field pattern analysis we found that the transmit waveshape was better described by a superposition of two spherical pulses

rather than a single spherical pulse. Based on this two-pulse model, we proposed new time delay calculations that extract signals associated with those pulses for data superposition. The new method, named coherent pixel-based (PB) beamforming, is shown to improve the image resolution significantly. Similar to SA imaging, however, most of the impact is on the lateral resolution which benefits directly from the large “synthesized” aperture. Extending the method to enhance the axial resolution is the primary purpose of this work.

In ultrasound imaging, the axial resolution can be improved by applying deconvolution after the spatiotemporal mapping. The beamformed 2-D RF data is assumed to be the output of a linear shift-invariant system, which can be modeled as a 2-D linear convolution between the *point-spread-function* (psf) and the scatterer field. Due to the variation of the beamshape with imaging depth, however, this assumption only holds within a small region. In [3], we implemented a 2-D Wiener filter using this approach, where the deconvolution is performed locally with the psf updated between individual *isoplanatic* patches. The results are showed to improve image resolution albeit at high computational cost. In addition, the method requires prior knowledge of the psf within each small region.

In this study, we propose a new way of combining the Wiener filter, as a pre-beamforming filter, with the coherent PB strategy. In particular, we extend the field pattern analysis in [4] and find that each received waveform can be deconvolved using the pulse echo generated from a single scatterer at the focus. This deconvolution helps suppress the interference of the excitation pulse and the spatial impulse response of transducer elements before the coherent PB can be applied. We then demonstrate the proposed method on two datasets, one is simulated with Field II [5] based on the parameters of the Verasonics Vantage 256 system and the other is acquired from

Fig. 1. The field pattern of a transmit beam, simulated after the Verasonics Vantage 256 system and L7-4 probe using Field II.

this system by scanning a tissue-mimicking phantom. The performance of the new method is evaluated and compared to the coherent PB algorithm based on the resolution of the generated images.

II. METHODS

In this section, we first review the field pattern analysis in [4] and the coherent PB beamforming in [2]. We then extend the analysis to form a Wiener filter, applied to the received waveforms before coherent PB beamforming.

A. Field Pattern Analysis and Coherent Pixel-Based Beamformer

We start with the expression for the pressure field arriving at imaging point P , given by [5]

$$p(\mathbf{x}_p, t) = \rho v(t) * \frac{\partial h_t(\mathbf{x}_p, t)}{\partial t}, \quad (1)$$

where $v(t)$ is a sinusoidal and uniform excitation on transmit, $*$ represents temporal convolution, $\mathbf{x}_p$ is the position vector of arbitrary imaging point P , ρ is the equilibrium density of the weakly-scattering medium, and $h_t(\mathbf{x}_p, t)$ is the transmit spatial impulse response.

By taking the transmit time delay into account, the transmit subaperture can be modeled as a continuous arc focused at point F , and $\partial h_t(\mathbf{x}_p, t) / \partial t$ can be approximated by [4]

$$\frac{\partial h_t(\mathbf{x}_p, t)}{\partial t} \approx \frac{c\beta_{\min}}{2\pi} \delta\left(t - \frac{R_{\min}}{c}\right) - \frac{c\beta_{\max}}{2\pi} \delta\left(t - \frac{R_{\max}}{c}\right), \quad (2)$$

where $R_{\max}$ and $R_{\min}$ are the maximal and minimal distances from the imaging point P to the aperture, $\beta_{\max}$ and $\beta_{\min}$ are angular coefficients associated with those distances, and c is the sound-speed. By submitting (2) into (1), we find that the transmit waveshape can be characterized by the superposition of two spherical pulses associated with $R_{\max}$ and $R_{\min}$.

This is supported by a simulation of the field pattern shown in Fig. 1. It is simulated using Field II [5], based on the parameters of the Verasonics Vantage 256 system and Philips L7-4 transducer (see the detailed system description in Sec III-A). In Fig. 1(a), we show how the limited angle α at focus

F [4] divides the field pattern into four regions, denoted (I) to (IV) counterclockwise. In each region, the transmit waveshape at imaging point P can be modeled as a superposition of two pulses, 180° out of phase (see Fig. 1(b)). The magnitudes of these pulses are varied among the regions. At the focus F , the pulses are merged into one and it has the biggest magnitude.

Based on this two-pulse model, we develop the coherent PB [2] with the focusing data given by

$$g(\mathbf{x}_p) = \sum_i^{N_t} \sum_j^{N_r} w_{ij} (c_1 r_{i,j}(\tau_{p,\min}) + c_2 r_{i,j}(\tau_{p,\max})), \quad (3)$$

where $r_{i,j}(t)$ is the RF data received on element j at transmission i , $\tau_{p,\min}$ and $\tau_{p,\max}$ are the time delays associated to the two pulses. Coefficients c_1 and c_2 are given by

$$c_1 = \frac{|\mathbf{x}_b - \mathbf{x}_p|}{|\mathbf{x}_b - \mathbf{x}_a|} \text{ and } c_2 = -\frac{|\mathbf{x}_a - \mathbf{x}_p|}{|\mathbf{x}_a - \mathbf{x}_b|}, \quad (4)$$

where $\mathbf{x}_a$, $\mathbf{x}_b$, $\mathbf{x}_p$ are the position vectors of points A, B, and P (see Fig. 1(a)). The minus sign in c_2 indicates the phase inversion between the two pulses of the transmit waveshape [2].

B. Coherent Wiener-filter Pixel-Based Beamformer

The RF signal received by element j reflected from a single point P is given by [6]

$$r_{j,P}(t) = \frac{\rho}{2c^2} E_m(t) * \frac{\partial^3 v(t)}{\partial t^3} * f_m(\mathbf{x}_p) * h_{pe}(\mathbf{x}_p, \mathbf{x}_j, t), \quad (5)$$

where $*$ represents spatial convolution, $\mathbf{x}_j$ is the position vector of the element j , $E_m(t)$ is the electro-mechanical impulse response from transducer force to voltage during the reception, $f_m(\mathbf{x}_p)$ accounts for the inhomogeneity in the soft tissue, $h_{pe}(\mathbf{x}_p, \mathbf{x}_j, t)$ is the spatial pulse-echo impulse response.

By separating the $h_{pe}(\mathbf{x}_p, \mathbf{x}_j, t)$ into $h_t(\mathbf{x}_p, t) * h_r(\mathbf{x}_p, \mathbf{x}_j, t)$, the RF signal in (5) can be re-written as

$$r_{j,P}(t) = \frac{\rho}{2c^2} E_m(t) * \frac{\partial^2 v(t)}{\partial t^2} * f_m(\mathbf{x}_p) * \frac{\partial h_t(\mathbf{x}_p, t)}{\partial t} * h_r(\mathbf{x}_p, \mathbf{x}_j, t), \quad (6)$$

Fig. 2. Simulated images of 6 point scatterers and 1 anechoic cyst with a random background using 3 different beamformers: (a) Dynamic focusing, (b) Coherent PB, (c) CWF-PB. The two pixel-based images are generated using data from 32 transmits. All image are log-compressed and displayed with a dynamic range of 70 dB.

TABLE I
PERFORMANCE METRICS ON SIMULATION

Beamformer	FWHM(Lateral, mm)			FWHM(Axial, mm)			CR
	20mm	28mm	36mm	20mm	28mm	36mm	
Dynamic focusing	0.633	0.508	0.605	0.634	0.610	0.632	0.358
Coherent PB	0.351	0.389	0.352	0.619	0.607	0.624	0.451
CWF-PB	0.338	0.382	0.347	0.491	0.482	0.473	0.453

where $h_r(\mathbf{x}_p, \mathbf{x}_j, t)$ is the receive spatial impulse response and it is assumed to be independent of point P . At focus F , (2) can be simplified as

$$\frac{\partial h_t(\mathbf{x}_f, t)}{\partial t} \approx \frac{c\beta_0}{2\pi} \delta\left(t - \frac{R_0}{c}\right), \quad (7)$$

and thus the signal echoed from a point scatter at the focal point F is given by

$$r_{j,F}(t) = v_{pe}(t) *_{\frac{t}{c}} \left(t - \frac{R_0}{c}\right) *_{\frac{t}{c}} h_r(\mathbf{x}_f, \mathbf{x}_j, t), \quad (8)$$

where $v_{pe}(t) \cong E_m(t) * \partial^2 v(t) / \partial t^2$, $\mathbf{x}_f$ is the position vector of point F , and R_0 represents the radius of the subaperture.

By assuming $h_r(\mathbf{x}_p, \mathbf{x}_j, t)$ has the same shape as $h_r(\mathbf{x}_p, \mathbf{x}_j, t)$, $r_{j,F}(t)$ in (2) can be used as a kernel to deconvolve the entire $r_j(t)$ in (6). This helps to suppress the blurring effects caused by $v_{pe}(t)$ and $h_r(\mathbf{x}_p, \mathbf{x}_j, t)$. The goal of this deconvolution is to recover the sharpness of the two pulses in $\partial h_t(\mathbf{x}_p, t) / \partial t$. Thus, the coherent PB beamforming calculation should be performed afterwards to extract the high-energy signals relating to the superposition of focusing data. We named the new strategy *coherent Wiener-filter pixel-based* (CWF-PB) beamformer.

III. EXPERIMENTS AND RESULTS

A. Data Acquisition and Evaluation Metrics

We demonstrate our proposed method on data simulated with Field II and that acquired in a phantom study using a Verasonics Vantage-256 system (Verasonics, Kirkland, WA, USA) and a linear array transducer (L7-4, Philips). The parameters of the simulation are extracted from the Verasonics system. The transducer has 128 elements and a 5.2083 MHz center frequency. In each transmit and receive, 64 elements are activated and the scanning step is 0.298 mm. The transmit focal depth is 25 mm while the sampling frequency is 20.83 MHz.

We evaluate the beamformer performance based on the spatial resolution and imaging contrast of the generated images [7]. Specifically, the spatial resolution is evaluated using the

full width at half maximum (FWHM) in lateral and axial directions, while the imaging contrast is evaluated based on the contrast ratio (CR), given by [2]

$$CR = \frac{|I_{out} - I_{in}|}{\sqrt{I_{out}^2 + I_{in}^2}}, \quad (9)$$

where I_{in} and I_{out} are the log-compressed mean intensities inside and outside the lesion, respectively.

B. Simulation

We first evaluate the beamformers on the simulated data generated from 6 point scatterers and 1 anechoic cyst embedded in a random background. The point scatterers are divided into three pairs, located at 3 different depths of 20 mm, 28 mm, and 36 mm. At each depth, the distance between the two points is 2.5 mm, 3.5 mm, and 4.5 mm, respectively. The anechoic cyst is centered at (6, 30) mm with a radius of 3.5 mm.

Figs. 2 shows the images generated using three different beamformers, including dynamic focusing, coherent PB, and CWF-PB. Comparing the coherent PB to the dynamic focusing (Figs. 2(b) versus (a)), we can observe significant improvements in the speckle representation and lateral resolution. This shows the good image resolution of coherent PB. Similar good performance is also observed with the CWF-PB in Fig. 2(c). Particularly, the lateral resolution is still maintained while the axial resolution is improved clearly through the use of the pre-beamformed Wiener filter.

To quantify the improvements offered by each beamformer, we measure the lateral and axial FWHM and CR for each of them at 20 mm, 28 mm, 36 mm and summarize the results in Table I. CWF-PB shows the best resolution, especially on the axial resolution. The lateral resolution of CWF-PB is similar to that of coherent PB, which maintains the good performance of coherent PB. Regarding the imaging contrast, the CRs calculated on coherent PB and CWF-PB images are almost the same and higher than that obtained with the dynamic focusing.

Fig. 3. Simulated images of 6 point scatterers and 1 anechoic cyst embedded in a random background using 3 different beamformers: (a) Dynamic focusing, (b) Coherent PB, (c) CWF-PB. All pixel-based images are generated with data from 32 transmits. They are log-compressed and displayed with a dynamic range of 70 dB.

TABLE II
PERFORMANCE METRICS ON PHANTOM STUDY

Beamformer	FWHM(Lateral, mm)			FWHM(Axial, mm)			CR
	A	B	C	A	B	C	
Dynamic focusing	0.398	0.565	0.670	0.453	0.443	0.416	0.120
Coherent PB	0.395	0.398	0.418	0.447	0.423	0.416	0.134
CWF-PB	0.357	0.363	0.404	0.398	0.354	0.345	0.146

C. Phantom Study

The experimental data is acquired using the Verasonics system by scanning a Multi-Purpose Multi-Tissue Ultrasound Phantom (Model 040, CIRS Inc.). The scanned region contains 7 wire targets whose radius is $40\mu\text{m}$ and one hyperechoic lesion with a 4 mm radius positioned at a depth of 30 mm. In this experimental study, we are still using the kernel from our simulation to form the pre-beamforming Wiener filter.

The imaging results are showed in Figs. 3. Similar to the simulation, the CWF-PB image (in Fig. 3(c)) offers a significant improvement in both lateral and axial resolutions compared to the dynamic focusing (in Fig. 3(a)), as well as good imaging of the speckle. The lateral resolution and CR of the CWF-PB are on par with that of the coherent PB (Fig. 3(b)). Table II shows the measured metrics, including lateral and axial FWHMs at points A, B, and C marked in Fig. 3(a). In the table, the dynamic focusing image is showed to have the lowest image quality while that with the CWF-PB is highest. For imaging contrast, the CWF-PB has the CR higher than those measured on both coherent PB and dynamic focusing images. These show the robustness of our proposed method toward experimental data.

IV. CONCLUSIONS

In this paper, we have proposed a new Wiener filter that aims to improve the axial resolution of medical ultrasound images. The filter is developed using an extension of our field pattern analysis. It is applied to the received waveforms before we perform coherent PB beamforming. In both simulated and experimental data, the new method is showed to offer significant improvements in axial resolution over the coherent PB while still maintaining the advantages of the coherent PB in the lateral resolution and imaging contrast. In the next step of this study, we will apply the new strategy to the *in-vivo* data to verify its potential for use in clinical settings.

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