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Improvement of range spatial resolution of medical ultrasound imaging by element-domain signal processing

Hideyuki Hasegawa

Graduate School of Science and Engineering for Research, University of Toyama, Toyama 930-8555, Japan

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1. Introduction

Ultrasonic imaging is widely used in clinical situations for the diagnosis of various diseases. It enables the noninvasive and real-time observation of biological tissues in vivo. The image quality in medical ultrasonic imaging is especially important to ensure an accurate diagnosis. The spatial resolution is one of the important factors determining the ultrasound image quality. At a certain ultrasonic frequency, the physical parameter determining the lateral spatial resolution is the aperture size. However, the lateral spatial resolution can also be improved by various signal processing techniques applied to ultrasonic echo signals received by individual transducer elements in an ultrasonic probe. In conventional ultrasonic imaging, element echo signals are delayed on the basis of geometrical information on the receiving focal point and summed to obtain beamformed radio-frequency (RF) signals. Such a procedure is called the delay-and-sum (DAS) beamforming. To improve the lateral spatial resolution, multiple compound imaging2–6 and synthetic aperture ultrasound imaging7–11 have been introduced. However, those methods require several transmissions to compound several images obtained with steered ultrasound beams, and the imaging frame rate is degraded.

To improve the lateral spatial resolution without compounding, several researchers showed that the lateral spatial resolution can be improved significantly using the coherence among the element echo signals.12–14 Furthermore, adaptive beamforming, namely, minimum variance beamforming, was introduced in ultrasound imaging for the improvement of the lateral spatial resolution.12–15 Those techniques realize a lateral spatial resolution, which is far better than that is determined by the physical aperture size in conventional DAS beamforming.

On the other hand, the range spatial resolution depends on the ultrasonic pulse length, which is determined by the mechanical response of the transducer element. To improve the range spatial resolution, signal processing methods based on the inverse filtering16–19 or frequency domain processing20,21 of beamformed ultrasonic echo signals have been proposed. Those methods basically increase the frequency bandwidth of the beamformed RF signal. As a result, the ultrasonic pulse length is reduced.

In the present study, methods that do not require the frequency analysis of ultrasonic echoes were proposed and applied to ultrasonic echo signals received by individual transducer elements. In conventional ultrasonic imaging using focused transmit beams, the above-mentioned methods for beamformed RF signals are feasible and the processing of element signals is not preferred because about 100 transmissions are required to construct an ultrasonic image and many element signals (72 elements in the present study) should be processed in each transmission. On the other hand, methods of processing element signals are feasible in high-frame-rate imaging because much fewer transmissions are necessary to construct an ultrasonic image. High-frame-rate ultrasound realizes a very high temporal resolution of more than 1,000 frames per second, and it is an emerging technique for measuring shear wave propagation22–24 and cardiovascular dynamics,25–27 and for monitoring high intensity focused ultrasound (HIFU) treatments.28–30 Thus, the image quality in high-frame-rate ultrasound should be improved. In the present study, the feasibilities of the proposed methods were evaluated in phantom experiments and the proposed methods were also applied to the in vivo measurement of a human carotid artery.

2. Principles

2.1 Discrete signal model

In the present study, range maximum likelihood (r-ML) and range multiple signal classification (r-MUSIC) filters, which were applied to the ultrasonic echo signal received by each transducer element, were proposed. In this section, the discrete signal model of the element echo signal is described.

Let us define the vector \( s \) of the sampled echo signal received by a transducer element as

\[
\mathbf{s} = (s_0 \quad s_1 \quad s_2 \quad \cdots \quad s_{N-1})^T,
\]

where \( N \) and the superscript \( T \) denote the number of sample points and the transpose, respectively, and \( s_n (n = 0, 1, 2, \ldots, N - 1) \) is the sampled element echo signal. Also, the transmit–receive response \( \mathbf{h} \) of the transducer element is defined as

\[
\mathbf{h} = (h_0 \quad h_1 \quad h_2 \quad \cdots \quad h_{K-1})^T,
\]

where \( K \) is the number of sampled responses.
In the present study, it was assumed that there is one point scatterer at each sampled point, and that \( M \) scatterers contribute to produce the echo \( s_n = (s_n, s_{n+1}, s_{n+2}, \ldots, s_{n+M-1})^T \) beginning from the \( n \)-th sampled point. Such a discrete signal model can be described as
\[
\hat{s}_n = Hz + n, \tag{3}
\]
where \( n \) is the noise vector and \( z \) represents the scattering strength of the scatterer at each sampled point defined as
\[
z = (z_0, z_1, z_2, \ldots, z_{M-1})^T. \tag{4}
\]
The \( K \times M \) matrix \( H \) is defined as follows:
\[
H = \begin{pmatrix}
h_0 & 0 & 0 & \cdots & 0 \\
h_1 & h_0 & 0 & \cdots & 0 \\
h_2 & h_1 & h_0 & \cdots & 0 \\
\vdots & \vdots & \vdots & \ddots & \vdots \\
h_{M-1} & h_{M-2} & h_{M-3} & \cdots & h_0 \\
h_M & h_{M-1} & h_{M-2} & \cdots & h_1 \\
\vdots & \vdots & \vdots & \ddots & \vdots \\
h_{K-1} & h_{K-2} & h_{K-3} & \cdots & h_{K-M-1}
\end{pmatrix}. \tag{5}
\]

### 2.2 Range maximum likelihood filter

On the basis of Eq. (3), the likelihood \( p_n \) of the element signal \( s_n \) is obtained, by assuming the noise \( n \) as Gaussian, as
\[
p_n = \frac{1}{\text{det}(\pi K)} \exp[-(s_n - Hz)^H K^{-1}(s_n - Hz)]. \tag{6}
\]
where the superscript \( H \) represents the Hermitian operator and \( K \) is the covariance matrix of the element signal \( s_n \) expressed as
\[
K = E[ss^H]. \tag{7}
\]
where \( E[\cdot] \) represents the expectation.

By considering the natural logarithm of Eq. (7), the logarithmic likelihood \( L_n \) is expressed as
\[
L_n = \ln p_n = -\log \text{det}(\pi K) - (s_n - Hz)^H K^{-1}(s_n - Hz). \tag{8}
\]
The scattering strength \( \hat{z} \), which maximizes the likelihood \( p_n \), is determined by setting the partial derivative of Eq. (8) with respect to \( z^H \) at zero.
\[
\frac{\partial L_n}{\partial z^H} = H^H K^{-1} (s_n - Hz) = 0. \tag{9}
\]
The estimated scattering strength \( \hat{z} \) is expressed as
\[
\hat{z} = H^H K^{-1} H^H (s_n - Hz) = K^{-1} H^H (s_n - Hz). \tag{10}
\]
Finally, the filtered element signal \( \hat{s}_n \) is obtained as \( \hat{s}_n = z_0 \), and the filtered element signal vector \( \hat{s} \) is obtained as \( \hat{s} = (\hat{s}_0, \hat{s}_1, \hat{s}_2, \ldots, \hat{s}_{M-1})^T \).

### 2.3 Range multiple signal classification filter

The multiple signal classification (MUSIC) method is used to create a sharp lateral directivity in the estimation of the direction of arrival\(^{37}\) and radar imaging.\(^{38}\) In the present study, the MUSIC method was used for the improvement of the range resolution in ultrasonic imaging.

![Fig. 1. Block diagram of beamforming procedure. (a) Conventional DAS beamforming. (b) Beamforming procedure used in the present study.]

By referring to Eqs. (3) and (7), the covariance matrix \( K \) of the element signal \( s_n \) is expressed as
\[
K = HzHz^H + nn^H = HQHz^H + \sigma^2 I. \tag{11}
\]
where \( \sigma^2 \) and \( I \) are the variance of noise and the identity matrix, respectively, and \( Q = zz^H \).

The proposed filter, namely, the r-MUSIC filter, estimates the eigenvalues \( \lambda_i \) (\( i = 0, 1, 2, \ldots, M - 1 \)) and the corresponding eigenvectors \( e_i \) (\( i = 0, 1, 2, \ldots, M - 1 \) of \( K \)), where \( \lambda_0 \geq \lambda_1 \geq \cdots \geq \lambda_{K-1} \). The eigenvectors \( e_0, e_1, \ldots, e_{M-1} \) are the signal eigenvectors, and the eigenvectors \( e_{M}, e_{M+1}, \ldots, e_{K-1} \) are the noise eigenvectors because it is assumed that there are \( M \) scatterers.

The echo signal from the \( j \)-th scatter \( g_j \) (\( j = 0, 1, \ldots, M - 1 \)), which corresponds to the column vector of \( H \), is orthogonal to the noise eigenvectors \( e_{M}, e_{M+1}, \ldots, e_{K-1} \). Thus,
\[
e_i^H g_j = 0 \tag{12}
\]
(\( i = M, M + 1, \ldots, K - 1; j = 0, 1, \ldots, M - 1 \)).

The output \( P(g_j) \) of the r-MUSIC filter is expressed as
\[
P(g_j) = \frac{g_j^H g_j}{\sum_{i=M}^{K-1} |e_i^H g_j|^2}. \tag{13}
\]
The filtered element signal \( \hat{s}_n \) is obtained as \( \hat{s}_n = P(g_j) \).

### 2.4 Receive beamforming

In the conventional DAS beamforming illustrated in Fig. 1(a), the echo signals from the outside of the receiving focal point are suppressed because the element signals have the phase information and the phases of the off-axis element signals are not aligned in the delay-and-sum procedure. However, the outputs of the proposed r-ML and r-MUSIC filters are the scattering strength, and the phase information of the original echo signal is removed. Therefore, the lateral resolution would be low when DAS beamforming is applied to the filtered element signals obtained using Eqs. (10) and (13). In the present study, the lateral directivity is created by multiplying the output of the conventional DAS beamformer with the unfiltered element signals to that with the filtered element signals, as illustrated in Fig. 1(b).
shown in Fig. 2 was used as the transmit 15-µm stainless wire. In the present study, the waveform contributing to the production of the echo signal respectively. In the present study, the number of scatterers applied to the element echo signals from the 15-µm stainless wire. The blue and green lines in Fig. 3(a) show the response \( h \).

The transmit–receive sequence is described in Ref. 28. In the present study, the number of emissions of plane waves for the creation of one image frame was set at 4, and each plane wave was emitted using 96 transducer elements. In each transmission, 24 focused receiving beams were created at intervals of 0.1 mm, and the aperture used to create one focused receiving beam consisted of 72 elements. Consequently, one image frame consisting of \( 24 \times 4 = 96 \) focused receiving beams was obtained by four emissions of plane waves.

3. Experimental results

3.1 Experiment using a fine wire

In the present study, an echo from a 15-µm stainless wire placed in water was measured to obtain the transmit–receive response \( h \). Figure 2 shows the measured RF echo from the 15-µm stainless wire. In the present study, the waveform shown in Fig. 2 was used as the transmit–receive response \( h \) \((K = 25)\).

Then, the proposed r-ML and r-MUSIC filters were applied to the element echo signals from the 15-µm stainless wire. The blue and green lines in Fig. 3(a) show the unfiltered (original) and r-ML filtered element signals, respectively. In the present study, the number of scatterers contributing to the production of the echo signal \( s_0 \) was assumed to be one \((M = 1)\) both in r-ML and r-MUSIC filtering. Therefore, in the present study, all the experimental data were analyzed by setting \( M = 1 \). The proposed r-ML and r-MUSIC filters were applied to a \( K \)-length sampled signal starting from the first sampled point of an \( N \)-length element signal to obtain the first sampled point of the filtered signal. Then, the same procedure was applied to a \( K \)-length signal starting from the second sampled point to obtain the second filtered sampled point. This procedure was repeated \((N - K)\) times to obtain all the sampled filtered signals. Other element signals were processed similarly.

As can be seen in the estimate of the r-ML filter, there are slight decays around the steep peak due to the slight correlation between the element signal \( s_0 \) and the transmit–receive response \( h \) because the transmit–receive response has a finite length. To suppress such decays, the absolute value of the derivative of the estimate shown in Fig. 3(a) was used as the final estimate of the r-ML filter, as shown by the green line in Fig. 3(b).

The blue and green lines in Fig. 4(a) show the unfiltered (original) and r-MUSIC filtered element signals, respectively. As in r-ML filtering, a very narrow and steep peak could be obtained by the proposed r-MUSIC filter. The estimate of the proposed r-MUSIC filter shown in Fig. 4(a) contains a small bias. Such a bias is caused by noise, which cannot be represented by the first eigenvector corresponding to the first eigenvalue \( \lambda_0 \). Therefore, in the present study, the bias of the r-MUSIC estimate was removed as shown by the green line in Fig. 4(b).

Figures 5(a)–5(c) show the B-mode images of the 15-µm stainless wire obtained without filtering, with the proposed r-ML filter, and with the proposed r-MUSIC filter, respec-

![Fig. 2](Image 61x691 to 276x780)

Fig. 2. (Color online) Waveform of echo from 15-mm stainless wire received by a transducer element.

![Fig. 3](Image 322x348 to 532x528)

Fig. 3. (Color online) Original and filtered element signals. (a) Original element echo signal (blue line) and r-ML estimate (green line). (b) Original element echo signal (blue line) and absolute value of derivative of r-ML estimate (green line).

![Fig. 4](Image 322x348 to 532x528)

Fig. 4. (Color online) Original and filtered element signals. (a) Original element echo signal (blue line) and r-MUSIC estimate (green line). (b) Original element echo signal (blue line) and bias-removed r-MUSIC estimate (green line).
tively. As can be seen in Fig. 5, the axial spatial resolution was significantly improved by the proposed r-ML and r-MUSIC filters. Figures 6(a)–6(c) show the beamformed RF signals obtained in the central lines in Figs. 5(a)–5(c), respectively, and Fig. 7 shows the corresponding axial amplitude profiles. From the amplitude profiles shown in Fig. 7, the axial spatial resolutions, which were defined using the widths at half maxima of the amplitude profiles, were evaluated to be 0.211 mm (conventional), 0.081 mm (r-ML), and 0.084 mm (r-MUSIC).

### 3.2 Experiment using phantom

In the previous section, the transmit–receive response, which was obtained from the element echo signal from the 15-µm stainless wire, was used to filter the same data. In the present study, the performance of the proposed methods was evaluated in another phantom experiment. The proposed methods with the transmit–receive response obtained using the 15-µm stainless wire were applied to the element echo signals from another phantom (CIRS 040GSE).

Figures 8(a)–8(c) show the B-mode images of the phantom obtained without filtering, with the proposed r-ML filter, and with the proposed r-MUSIC filter, respectively. In Fig. 8, the dynamic ranges (DRs) of the B-mode images are controlled so that such B-mode images can be observed similarly. As can be seen in Fig. 8, the r-MUSIC filter realizes better penetration than the r-ML filter.

Figure 9 shows the axial echo amplitude profiles obtained in the lines, which are located at the strings in B-mode images shown in Fig. 8.
by the proposed methods even when the transmit–receive response was used in the filtering of a different target. The axial widths at half maxima of the amplitude profiles in Fig. 9 were evaluated to be 0.209 mm (conventional), 0.086 mm (r-ML), 0.094 mm (r-MUSIC).

3.3 In vivo imaging of human carotid artery

The feasibility of the proposed methods was evaluated by the in vivo imaging of a human carotid artery. Figures 10(a)–10(c) show the B-mode images of the carotid artery obtained without filtering, with the proposed r-ML filter, and the proposed r-MUSIC filter, respectively. In Fig. 10, the DRs of the B-mode images are controlled so that such B-mode images can be observed similarly. As can be seen in Fig. 10, the echo from the lumen-intima interface of the posterior wall becomes sharper in the B-mode images obtained using the proposed r-ML and r-MUSIC filters [Figs. 10(b) and 10(c)] than in that obtained without filtering [Fig. 10(a)]. Also, the continuity of the echo from the lumen-intima interface of the anterior wall, which is indicated by the red arrows in Fig. 10(a), is significantly improved. The reason considered was that the destructive interference among echoes could be reduced by increasing the axial spatial resolution. Furthermore, the undesired echoes in the lumen, which are indicated by the cyan arrow in Fig. 10(a), can be suppressed by the proposed r-ML filter. As demonstrated by these results, the proposed method is also feasible in imaging biological tissues.

4. Discussion

In the present study, two methods based on the ML and MUSIC estimators were proposed for the improvement of the axial spatial resolution in high-frame-rate ultrasound imaging. The proposed r-ML and r-MUSIC filters require the transmit–receive response of the transducer element. In the present study, the echo from the 15-µm stainless wire placed in water was used as the transmit–receive response. The proposed methods improved the axial spatial resolution significantly when they were applied to the same echo data from the 15-µm stainless wire. The improved axial spatial resolution slightly degraded when the proposed method was applied to another phantom. The difference between the attenuation properties of the propagation media could be considered as a major reason. The attenuation coefficient of the tissue-mimicking material in the phantom was significantly larger than that of water. The transmit–receive response was supposed to be changed by the difference in the frequency-dependent attenuation of the propagation media. In future works, the performance of the proposed filters should be improved by considering the attenuation property of the propagation medium.

In the present study, the regional echo signal was modeled by a single scatterer because the largest eigenvalue was larger than the second largest eigenvalue by more than 50 dB, which is a typical dynamic range for displaying a B-mode image. The covariance matrix was obtained from the echo signal during a short period corresponding to one ultrasonic pulse length. Therefore, the echo signal was considered to be modeled by that from one scatterer.

The estimates obtained by the proposed r-ML and r-MUSIC filters correspond to the scattering strengths of scatterers. An ultrasonic echo received by each transducer element is the convolution of the scattering strength and the transmit–receive response of an element, which is an alternative current (AC) component. Out-of-focus echo signals received by transducer elements also become an AC component across the array even after delay compensation. An AC component is suppressed by the delay-and-sum procedure, and the lateral directivity is created. On the other hand, the scattering strengths obtained by the proposed r-ML and r-MUSIC filters contain large direct current (DC) components. A direct current component is not suppressed by the delay-and-sum procedure, and the lateral directivity becomes unsharp. Figures 11(a) and 11(b) show the B-mode images obtained by the delay-and-sum procedure applied only to the outputs of the proposed r-ML and r-MUSIC filters, respectively. The lateral resolution of the B-mode images in Fig. 11 becomes worse than that of the conventional B-mode image shown in Fig. 5(a). Therefore, in the present study, the lateral directivity was created by the delay-and-sum procedure applied to ultrasonic echoes received by transducer elements.
An ultrasound image can be created by much fewer transmissions in high-frame-rate imaging, i.e., 4 in the present study, than in conventional linear scanning. Therefore, the filtering applied to the element signals is feasible. In the present study, the number of elements in the receiving aperture was 96, and $4 \times 96$ element signals must be filtered. However, conventional linear scanning requires a much larger number of transmissions to create an ultrasonic image, and it is computationally difficult to filter the element signals. In conventional linear scanning, it would be better to apply the proposed methods to the beamformed RF signal to reduce the computational load. However, the transmit-receive response of the beamformed RF signal depends on not only the attenuation property of the propagation medium but also the focusing effect, and such an effect should be considered. As described above, although some factors should be taken into account for further improvement of the performance of the proposed methods, the proposed methods do not require the Fourier or Hilbert transform to obtain the frequency spectrum or analytic signal of the ultrasonic echo and can be applied to real ultrasound echo signals easily. Such a property of the proposed methods would be beneficial for high-resolution ultrasound imaging.

In the present study, two filters based on the ML and MUSIC methods were proposed, and the performance characteristics of those methods were examined and compared. Both the proposed r-ML and r-MUSIC filters realized a significantly better range spatial resolution than the conventional method. However, the proposed r-ML and r-MUSIC filters showed some different characteristics, i.e., the penetration and suppression of undesired echoes in the lumen of the carotid artery shown in Fig. 10. The proposed r-ML filter was considered to suppress echoes that poorly correlated with the reference signal. The waveform of an echo from a deeper region is more deformed than that from a shallow region owing to the frequency-dependent attenuation of ultrasound. Therefore, the penetration of the proposed r-ML filter was not good because echoes from a deep region were suppressed by the deformation of the echo waveform. Similarly, the undesired echoes in the lumen of the carotid artery were suppressed by the proposed r-ML filter because of the frequency-dependent attenuation during multiple reflections. Such an additional effect of the proposed r-ML filter will be investigated in our future work.

5. Conclusions

The range spatial resolution is one of the important metrics determining the ultrasound image quality. In the present study, two element-domain signal processing methods were proposed for high-frame-rate ultrasonic imaging. The results of the phantom experiments showed that the proposed methods improved the range spatial resolution significantly. Also, the result of the in vivo measurement of a human carotid artery showed a significant improvement of the image quality. The proposed methods are particularly suitable for high-frame-rate imaging, in which an ultrasonic image is created by much fewer ultrasound transmissions than conventional linear scanning.

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