Singular value decomposition filtering in high-frame-rate cardiac vector flow imaging

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ABSTRACT

Dysfunction of the left ventricle (LV) weakens the cardiac function and affects the physical activity. Echocardiography has been used to visualize the blood flow dynamics and to evaluate the cardiac function. However, the signal processing to suppress the clutter signals should be employed. In this study, we employed the singular value decomposition (SVD) clutter filtering to obtain the cardiac blood speckle images. We also employed the adaptive thresholding metric to determine the proper cutoff values at each phase during the cardiac cycle. Moreover, we employed a depth-dependent SVD clutter filter for more accurate estimation of the cardiac blood echo signals. The 2D blood flow velocity vectors were estimated by applying the block matching method to obtained blood speckle images. The obtained results show that the proposed filter suppressed the clutter signals from left ventricular wall significantly, and the contrast-to-noise ratio (CNR) was improved from -0.5 dB to 13.8 dB by the proposed SVD clutter filtering.

Keywords:
Block matching method
Cardiac blood flow
Clutter filter
Singular value decomposition
Ultrasound imaging

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1. INTRODUCTION

Echocardiography has been used to evaluate the cardiac function. The color-Doppler imaging is a credible technique for visualization of blood flow [1] and used for visualizing of the blood flowing in and out from the cardiac cavity. In echocardiography, the color-Doppler imaging is also employed to visualize an abnormal blood flow such as mitral regurgitation [2]. In the color-Doppler imaging, the flow velocity component only in the axial direction is estimated, and the velocity component parallel to the axial direction cannot be obtained. In this sense, the color-Doppler image doesn’t stand for the true velocities.

The block matching method is able to estimate 2D velocities [3-6]. In this method, the motion of speckle patterns between successive frames is tracked and, thus, blood speckle images are required to obtain flow velocities. However, the intensity of the clutter echo is 40-60 dB higher than those originating from red blood cells [7]. Therefore, the signal processing to suppress the clutter signals should be employed. In our previous study, the high pass filter, which chooses the cut-off frequency adaptively based on the velocity of the heart wall [8], could visualize the blood B-mode images at a high-frame rate [9, 10].

Recently, the singular value decomposition (SVD) gains attention in research fields of image processing as a low-lank approximation method. The SVD-based filtering was applied to diagnostic imaging modalities [11, 12], and the SVD-based filter contributes to reduction of noise components contained in the diagnostic images. In the field of the medical ultrasound, the SVD-based method was employed as
a clutter filter [13-16] and its performance is better than that of the conventional finite impulse response (FIR) filter [13, 17]. The application of the SVD clutter filter is currently limited to blood flow in small vessels [13-15]. To assess the feasibility of the SVD filter in cardiac blood flow imaging, in this study, we applied the SVD clutter filter to echo signals the received from the human heart. In clutter filtering of ultrasound echo from the heart wall, the heart wall exhibits remarkable movements during a cardiac cycle. For this reason, the fixed cutoff values cannot be adopted in the cardiac blood flow, and the cutoff values in a clutter filter should be altered during the cardiac cycle adaptively based on the information of the clutter motion as described in our previous study [8]. To solve such a problem, we employed the adaptive thresholding metrics in the SVD clutter filter.

In the cardiac blood flow imaging, the optimum cutoff values were different at the positions of the echo data. Therefore, it is desirable to apply the clutter filters to spatially segmented ultrasound data. In this study, we applied a depth-dependent SVD clutter filter to obtain the cardiac blood echo signals. Also, an envelope-based block matching method was applied to the cardiac blood speckle images to obtain the 2D velocity vector maps. The performance of the proposed method was evaluated in the in vivo experiment on the human heart.

2. RESEARCH METHOD

2.1. Experimental setup

A phased array probe at a center frequency of 3 MHz was used in the in vivo experiment. One diverging beam was transmitted using the transducer elements of the phased array probe [18, 19]. Element echo signals were received and applied the delay-and-sum (DAS) beamforming. The receiving focal points were aligned in the Cartesian coordinate to avoid a redundant effect due to non-constant lateral sampling interval in the polar coordinate [19]. As a result, beamformed RF signals along 241 scan lines were obtained. Under such conditions, a frame rate at 6250 Hz was achieved.

2.2. SVD clutter filtering method

A Let us define the beamformed RF signals \( s(x, z, t) \), where \( x, z, \) and \( t \) represent the lateral position, depth position, and time, respectively. The two-dimensional matrix \( S \) is created by rearranging the beamformed RF data \( s(x, z, t) \) to the Casorati matrix with a size \( (n_x \times n_z, n_t) \), where \( n_x, n_z, \) and \( n_t \) is the number of pixels in the lateral and depth direction, and packet size of filter, respectively. By applying SVD, the matrix \( S \) can be decomposed into a production of three matrices as:

\[
S = UV\Sigma V^T
\]  

(1)

where \( U \) and \( V \) are matrices composed of spatial and temporal singular vectors, \( \Sigma \) is a diagonal matrix composed of the singular values arranged in a descending order. The superscript \( .^T \) represents transpose. The singular values obtained by such a procedure corresponds to energy of the received ultrasound signals, and the low-order, intermediate-order, and high-order singular components are assumed to be related to the clutter, blood, and noise signals, respectively. The clutter filter was accomplished by setting singular values corresponding to clutter signals to zero as:

\[
S' = U\Sigma' V^T
\]  

(2)

where \( \Sigma' \) is a diagonal matrix obtained by substituting singular values corresponding to clutter signal with zero. The filtered signals were obtained by reshaping \( S' \) to the original size. Figure 1 shows a schematic of the SVD clutter filtering.

2.3. Thresholding of SVD clutter filter

In this section, the thresholding metric for creating the matrix \( \Sigma' \) was introduced [15, 20]. In the SVD clutter filtering, the low-order components, which are assumed to be related to the clutter signals, were removed. Therefore, the boundaries between the clutter and blood signal components should be determined. As described in Section 1, in cardiac blood flow imaging, a cutoff threshold value of the clutter filter should be adaptively because the velocity of the heart wall changes rapidly during the cardiac cycle.
Singular value decomposition filtering in high-frame-rate cardiac vector flow imaging (M. Mozumi)

In this study, the blood subspace boundaries were determined based on the similarity of the spatial matrix $U$. The similarity between $n$-th and $m$-th column vectors of $U$ was calculated as:

$$C(n, m) = \frac{1}{n_z n_x} \sum_{x,z} \left( |U_n(x, z)| - |U_m(x, z)| \right) \times \left( |U_n(x, z)| - |U_m(x, z)| \right)$$ \hspace{1cm} (3)

where $U_n(x, z)$ is the $n$-th spatial singular vector, and $\bar{\cdot}$ stands for averaging. The spatial singular vector corresponds to B-mode images. The two spatial singular bases, which belong to the same subspace, have high spatial correlation. Therefore, we determined the clutter and blood signal subspaces using the spatial similarity matrix $C$. Let us define the parameter matrix $a_{a,b}$ expressed as $[15, 20]$.

$$a_{a,b} = \begin{cases} 1 & \text{if } (n, m) \in [1, a]^2 \\ 1 & \text{if } (n, m) \in [a, b]^2 \\ 0 & \text{otherwise} \end{cases}$$ \hspace{1cm} (4)

The clutter and tissue and noise subspace boundaries, $a$ and $b$, were searched so that the following normalized correlation coefficient $\chi_N$ was maximized:

$$\chi_N = \frac{\chi(C, a_{a,b})}{\sqrt{\chi(C, C) \chi(a_{a,b}, a_{a,b})}}$$ \hspace{1cm} (5)

where

$$\chi(C, a_{a,b}) = \frac{1}{n_z n_x} \sum_{x,z} \left( |C(x, z)| - |C(x, z)| \right) \times \left( |a_{a,b}(x, z)| - |a_{a,b}(x, z)| \right)$$ \hspace{1cm} (6)

Figure 2 shows a schematic of an adaptive thresholding metric in this study. In the clutter subspace and in the blood subspace, the spatial bases in the same subspace were highly correlated. In Figure 2, an area depicted as the red square was assigned to be the blood subspace. Then, the diagonal matrix $\Sigma'$ was created so that the singular values except for the blood signal components were set at zero.
2.4. Depth-dependent SVD clutter filter

In Section 2.2, the SVD clutter filter is applied to the whole region of the received ultrasound data. However, in the cardiac blood flow imaging, the optimum cutoff values were different at the spatial positions of the echo data [8]. Therefore, it is desirable to apply the clutter filters to spatially segmented ultrasound data. In this study, we employed a depth-dependent SVD clutter filter to obtain the cardiac blood echo signals. As shown in Figure 3(b), regions of interest (ROI) were assigned by scanning a template window in the depth direction. The ultrasound data $M_i$ in the segmented regions were subjected to the SVD clutter filter and decomposed into a product of three matrix as:

$$M_i = U_i \cdot \Sigma_i \cdot V_i^T$$

(7)

where $i = 1, 2, ..., N_{ROI}$, and $N_{ROI}$ is the number of the ROIs. Blood signal components were extracted in each segmented data as:

$$M'_i = U_i \cdot \Sigma'_i \cdot V_i^T$$

(8)

where $M'_i$ is $i$-th the output signals of the SVD clutter filter. In this method, the blood signal components were estimated by the distribution of the singular values. To suppress the clutter signal components, we assigned the components with singular values of -55 dB lower than the maximum singular value as the blood signal components. Also, to suppress the noise components, the noise components were estimated using the slope of the distribution of the singular values [21].

Finally, we concatenated output signals from each SVD filter to form one B-mode image. In this study, pixels in two adjacent ROIs were overlapped to reduce spatial discontinuity of the final output data. The numbers of overlapped pixels of adjacent ROIs were set to be 95% of a block size in the depth direction of ROI. In this case, one B-mode image was formed by concatenating 5% of output signals from each SVD filter.

2.5. Estimation of the blood velocity vectors

In this study, 2D velocity vectors were estimated by applying the block matching algorithm to the blood speckle images (envelope signals) obtained with the clutter filtering. The normalized correlation function between successive blood speckle frames was calculated with the kernel size of (6.0 mm x 7.9 mm) in the lateral and depth directions. The obtained normalized correlations were averaged temporally to suppress undesired components and interpolated with the reconstructive interpolation to obtain the subsample displacements [22-25]. Blood velocity vectors were obtained from the interpolated correlation function maps and displayed on the blood speckle B-mode images.
2.6. In vivo experiment

The ultrasound echo signals from the left ventricle of a 27-year-old healthy volunteer were analyzed. The blood flow B-mode image were obtained by applying the SVD clutter filter to the beamformed RF signals. The B-mode image obtained with the SVD clutter filter was compared to that obtained with the FIR-based clutter filter, which chooses the cutoff frequency based on the velocity in the heart wall [8]. To evaluate the performance of the clutter filter, we measured the contrast-to-noise ratio (CNR) of the filtered images. CNR was expressed as:

$$NR = \frac{PW(s_{\text{blood}}) - PW(s_{\text{tissue}})}{\text{std}(s_{\text{tissue}})}$$

(9)

where $s_{\text{blood}}$ and $s_{\text{tissue}}$ are echo signals which correspond to blood and clutter (background), and $PW(s)$ denotes the power of the signal expressed as:

$$PW(s) = \frac{1}{n_xn_z} \sum_{x,z}|s(x,z)|^2$$

(10)

Also, $\text{std}(\cdot)$ represents the standard deviation.

3. RESULTS

Figure 4(a)-(c) show cardiac blood flow images at a systolic phase obtained without clutter filtering (original beamformed signals), with the FIR-based filtering, and with the SVD clutter filtering, respectively. In Figure 4(c), the undesirable clutter signals at a depth of 100 mm were more suppressed than those shown in Figure 4(b), whereas an appearance of the blood speckles were preserved. Regions of interest (ROIs) of the blood and clutter (background) were depicted in Figure 4(a) as red rectangles. The CNRs of B-mode images in Figure 4(b) and 4(c) became -5.8 and 12.3 dB.

Figure 4. B-mode image of left ventricular at a systole phase, (a) Original beamformed signals, (b) Cardiac blood flow images obtained with FIR-based filter, (c) With the SVD clutter filter, ROIs labeled with blood and tissue represent pixels corresponding to cardiac lumen and heart wall, respectively.

Figure 5(a)-(c) show cardiac blood flow images at a mid-diastolic phase obtained without clutter filtering, with the FIR-based filtering, and with the SVD clutter filtering, respectively. In Figure 5(c), signals which corresponding to the heart wall were also suppressed as compared with the B-mode image in Figure 4(b). In this case, the blood speckle was less affected, but the SVD clutter filter was capable to suppress the clutter signals. The CNRs of B-mode images in Figure 5(b) and 5(c) became -0.5 and 13.8 dB. Figure 6(b) and 6(c) show cardiac blood flow images in systolic phase obtained by the SVD clutter filter without and with the segmentation of ROI in depth direction.
Figure 5. B-mode image of left ventricular at a mid-diastole phase, (a) Original beamformed signals, (b) Cardiac blood flows image obtained with FIR-based filter, (c) With the SVD clutter filter

Figure 6. B-mode image of left ventricle in systole phase, (a) Original beamformed signals and obtained by SVD-based filter, (b) Without, (c) with the segmentation of ROI in depth direction

Figure 7(b) and 7(c) show cardiac blood flow images in diastolic phase obtained by the SVD clutter filter without and with the segmentation of ROI in depth direction. In those results, \( n_t \) is set to be 512. As shown in Figure 7(c), the contrast of echoes from blood cells is improved than Figure 7(b) by applying the depth-dependent SVD clutter filters to each ROIs. The CNRs of B-mode images in Figure 6(b) and 6(c) became 17.2 and 20.0 dB, and Figure 7(b) and 7(c) became 9.7 and 14.9 dB. Figure 8(a) and 8(b) show the 2D blood velocity vectors of the cardiac blood flow at a mid-diastolic phase obtained with the FIR-based filtering and with SVD clutter filtering. In Figure 8(b), vortex flows in the left ventricle were captured well in the proposed method.
Singular value decomposition filtering in high-frame-rate cardiac vector flow imaging (M. Mozumi)

4. CONCLUSION

Visualization of the blood flow vector is useful to evaluate behavior of the cardiac blood flow. In this study, we applied the SVD-based clutter filter to the ultrasound echo signals from the human heart to obtain the cardiac blood flow image. We also employed the adaptive thresholding metric to determine the proper cutoff values through the cardiac cycle. Moreover, we employed a depth-dependent SVD clutter filter for more accurate estimation of the cardiac blood echo signals. The experimental results show that the proposed filter suppressed the clutter signals from left ventricular wall, and the CNR evaluated using the blood speckle image was improved from -0.5 dB to 13.8 dB by the SVD clutter filtering.
ACKNOWLEDGEMENTS

This study was supported by JSPS KAKENHI Grant Number JP17H03276.

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Singular value decomposition filtering in high-frame-rate cardiac vector flow imaging (M. Mozumi)

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