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Relationship between singular value decomposition filter settings and image contrast in backscattering analysis of blood

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1. Introduction

In ultrafast imaging (UI) of blood flow, a clutter filter suppresses echoes from surrounding tissue such as vessel walls (clutter signal) and system-derived noise signals, enabling clear visualization of blood flow signals. The possibility of in vivo evaluation of erythrocyte aggregation has been investigated by analyzing the frequency dependence of backscatter coefficient (BSC) of the clutter-filtered echo data[1]. Because clutter filter settings are empirical in the previous study, an adaptive approach of filter design is necessary to perform stable backscatter coefficient analysis. This study confirms the effects of adaptive filter settings based on the image quality in two center frequency conditions of 7.5 and 12.5 MHz for the BSC analysis. Adaptive filter settings were validated in the dataset of blood flow phantom with various flow rates.

2 Method

2.1 Test data

Samples of pig erythrocytes diluted in saline to a hematocrit value of 3% or 40% were used for UI by steady-state circulation in the lumen of a phantom containing a clutter signal. The reason for the 3% hematocrit value is to make the hematocrit value closer to the ideal value for BSC without causing aggregation of red blood cells to be used as reference data for BSC. In addition, 40% hematocrit values were set to hematocrit values that were close to physiological values to provide data for analysis. The flow rates of blood flow were 10, 30, 50, 100, 150, 250, 350, 450, and 600 mL/min. The plane wave transmission sequence was adopted at two frequencies (7.5 and 12.5 MHz) excitation of the wideband linear probe developed in our previous study[2].

2.2 Singular value decomposition filter

A singular value decomposition (SVD) filter was applied to suppress the effect of clutter signals and to enhance the blood echo signal. Let us consider a two-dimensional matrix **A** [depth × azimuth (space), frame direction] transformed from a three-dimensional RF signal (depth, azimuth, frame direction) after delay-and-sum beamforming. The two-dimensional matrix **A** is represented by the product of three matrices by SVD.

$$\mathbf{A} = \mathbf{USV}^H. \tag{1}$$

Table. I Low-rank threshold of the SVD filter in the reference data. Two of center frequencies (7.5 and 12.5 MHz) dataset was separately filtered.

Flow rate [mL/min]	Frequency [MHz]	Threshold
0	7.5	3
	12.5	3

Table. II Low-rank threshold of the SVD filter at each flow rate in the analyzed data.

Flow rate [mL/min]	Frequency [MHz]	Low rank threshold
10	7.5	4
	12.5	4
30	7.5	4
	12.5	7
50	7.5	4
	12.5	6
100	7.5	7
	12.5	6
150	7.5	7
	12.5	7
250	7.5	8
	12.5	9
350	7.5	9
	12.5	12
450	7.5	11
	12.5	14
600	7.5	10
	12.5	13

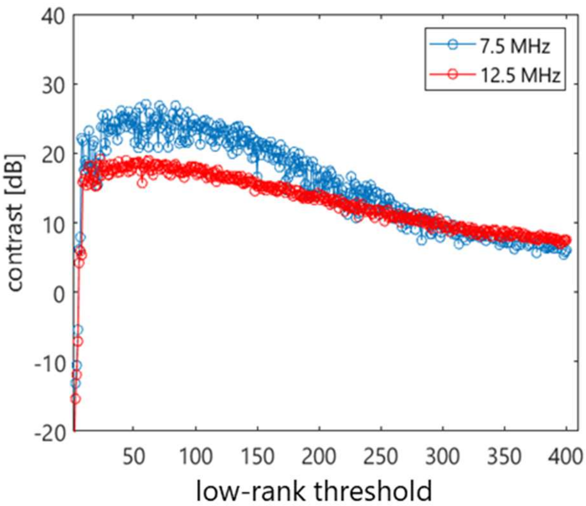

Fig. 1 Contrast results at a flow rate of 250 mL/min

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where $\mathbf{U}$ and $\mathbf{V}$ (H is the complex transpose) are matrices consisting of singular value vectors in space and time, and $\mathbf{S}$ is a body matrix with singular values in descending order in the diagonal terms. Based on the diagonal matrix of singular values, the low-rank component is assumed as clutter components, and their suppression is used to enhance the blood flow component[1]. The criteria of the threshold settings based on the singular value number, i.e., low-rank threshold, between the clutter and blood flow components was calculated by the following analysis of image contrast.

2.3 Contrast analysis

To evaluate image quality performance under arbitrary filter conditions, the contrast between the blood flow and clutter areas is analyzed. The contrast is expressed by the following equation.

$$contrast = 20\log_{10}\left(\frac{ROI_2}{ROI_1}\right), \qquad (2)$$

where ROI_1 and ROI_2 are the mean of the amplitude envelope of all frames in the region of interest in the clutter and blood flow part. Once the low-rank threshold was determined, the singular value numbers from 0 to low-rank threshold was suppressed in the SVD filter. Thus, a threshold of the high-rank threshold between blood flow and noise components was constantly set at 400. **Fig. 1** shows the result of contrast at an arbitrary low rank threshold at the flow rate of 250 mL/min. A first inflection point such as 8 in 7.5 MHz and 9 in 12.5 MHz dataset was manually selected. **Tables I and II** summarize the low-rank thresholds of the SVD filters at center frequencies of 7.5 and 12.5 MHz for the reference and analysis data. Overall, in both center frequencies, the low-rank threshold was slightly higher in the increase of flow rate.

2.4 Backscattering coefficient analysis

BSC is a physical quantity related to the size, density, and acoustic impedance ratio of the scatterer[3]. In this report, BSC is calculated by reference phantom method as following equation[4].

$$BSC(f) = \frac{P(f)}{P_{ref}(f)} \cdot \frac{A_{ref}(f)}{A(f)} \cdot BSC_{ref}(f), \qquad (3)$$

where P and P_{ref} are the average frequency power spectra of the analysis and reference data windowed to the blood flow portion, respectively. A and A_{ref} are the known attenuation characteristics of the analysis and reference data, respectively. For the calculated BSC, the linear approximation shown in Equation (4) was performed within the range of 5–10 MHz and 10–15 MHz for 7.5 and 12.5 MHz dataset.

$$10\log_{10}[BSC(f)] \approx n\log_{10}f + 10\log_{10}m. \qquad (4)$$

The slope n of the theoretical line of BSC following Rayleigh scattering is approximately 4[1].

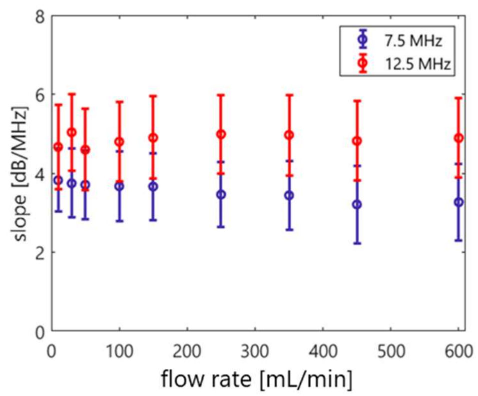

Fig. 2 Slope of the BSC in the different dataset of flow rate with adaptive threshold determined in Table II.

3. Result and Discussions

Fig. 2 shows the slope of the BSC at each center frequency and flow rate at the threshold determined in Table I. In the center frequency of 7.5 MHz, the average slope was decreased from 4.0 to 3.5 with the increase of the flow rate. In contrast, the average slope was increased from 4.0 to 4.5 for higher flow rate in the center frequency of 12.5 MHz. While it is assumed that the frequency dependence of the BSC is constant due to Newtonian fluid of erythrocyte suspended by saline, slight changes of the slope was confirmed at different flow rates at either frequency. In future works, we compared the results in constant thresholding of the SVD filter to this adaptive approach.

4. Conclusion

In this study, we confirmed the relationship between the singular value decomposition filter setting conditions and image contrast, and feedback for the BSC analysis of blood. As a result, the slope of the BSC was changed in the different flow rate under the conditions of the SVD filter based on the current image contrast. In the future, we will examine how fine changes in the threshold value on the low-rank side of the SVD filter affect the BSC results.

References

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