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Transcranial Doppler ultrasonography using spatial domain interferometry with Capon method: simulation study
Capon 法と空間領域干渉計法を用いた経頭蓋骨血流速度
推定方法の検討

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1. Background

Subarachnoid hemorrhage (SAH), bleeding inside the cranium, is one of the most important and severe conditions for neurosurgeons to treat. From 20% to 40% of SAH patients have poor outcomes. One of the main causes for the poor outcome is vasospasm that reduces the blood flow and induces serious conditions, e.g. brain infarct. The highest vasospasm risk period is from 3 to 14 days after SAH, therefore the dairy test for vasospasm is required.

Transcranial Doppler ultrasonography (TCD) is a non-invasive and repeatable test to detect the vasospasm. However, the low signal-to-noise ratio in the cranium deteriorates the accuracy of TCD. We have reported the effect of the moving target indicator (MTI) filter and the spatial domain interferometry (SDI) with the Capon method on the suppression of high-intensity interferences [1]. In this study, we investigate the temporal and range resolution of the proposed method.

2. Methods

The proposed method employs an MTI filter. The MTI filter subtracts the previous received signal from the received signal to suppress the echoes from stationary targets. However, the interference from the cranium has high echo-intensity and its time-varying component should not be negligible.

To suppress these time varying components, we apply the SDI method with the Capon method to the differential signals after MTI filtering. The Capon method is one of the adaptive beamforming techniques for high resolution imaging. The SDI method is a technique for estimating the echo intensity from a desired angle. The output of SDI $y_k(t)$ is given by:

$$y_k(t) = \mathbf{W}_k^{\mathrm{T}*}\mathbf{x}_k(t) \qquad (1)$$

$$\mathbf{x}_k(t) = [x_{k,1}(t) \quad \cdots \quad x_{k,L}(t)]^{\mathrm{T}}, \qquad (2)$$

where $x_{k,l}(t)$ is the l-th element of the differential signal at the k-th transmit event, $\mathbf{W}_k$ is a weighting vector, $*$ denotes the complex conjugate and T denotes transpose. The output power at a desired depth d is given by

$$P_{\mathrm{out}}(d) = \mathbf{W}^{\mathrm{T}*}\mathbf{R}\mathbf{W}, \qquad (3)$$

where $\mathbf{R}$ is the covariance matrix of the differential signals at the depth of d.

The desired signal from red blood cells is supposed to have low correlation with the interferences from undesired stationary targets. Therefore, the proposed method works without averaging sub-matrices in the diagonal direction. We can use two averaging processes to estimate the covariance matrix R; one is the averaging in the range direction, and the other is the temporal averaging. The (i,j)-th element of the covariance matrix $\mathbf{R}$ is expressed by

$$r_{ij} = \frac{1}{mT_{\mathrm{ave}}}(\sum_{k=1}^{m}\int_{T_{\mathrm{obs}}-T_{\mathrm{ave}}/2}^{T_{\mathrm{obs}}+T_{\mathrm{ave}}/2} x_{k,i}(t)^* x_{k,j}(t)\mathrm{d}t), \quad (4)$$

where m is the number of the temporal averaging, $cT_{\mathrm{obs}}/2$ is the desired range and $cT_{\mathrm{ave}}/2$ is the average length in the range direction, c is the sound velocity. Eq. (4) indicates that the temporal resolution is a trade-off for the spatial resolution in the proposed method.

The SDI method with the Capon method is given by:

$$\min P_{\mathrm{out}} = \mathbf{W}^{\mathrm{T}*}\mathbf{R}\mathbf{W} \quad \text{subject to } \mathbf{C}^{\mathrm{T}*}\mathbf{W} = 1, (5)$$

where $\mathbf{C}$ is a constrainment vector. When the desired signal arrives from the front of the probe, all components of $\mathbf{C}$ are equal to 1. The solution of Eq. (5) is given by

$$\mathbf{W}_{\mathrm{opt}} = \frac{H^*}{\mathbf{C}^{\mathrm{H}}(\mathbf{R}+\eta\mathbf{E})^{-1}\mathbf{C}}(\mathbf{R}+\eta\mathbf{E})^{-1}\mathbf{C}, \qquad (6)$$

where $\mathbf{W}_{\mathrm{opt}}$ is the optimal weighting vector determined by Eq. (5), H is a constrained value and its value is equal to 1 when the desired signal arrives from the front side of the probe. $\eta\mathbf{E}$ is a diagonal loading matrix used to stably obtain the inverse matrix $\mathbf{R}$. The output of the SDI method

with the Capon method $y_{optk}(t)$ is given by:

$$y_{optk}(t) = \mathbf{W}_{optk}^{T*} \mathbf{x}_k(t) \cdot \qquad (7)$$

We evaluated our proposed method using a 2-D ray-tracing model simulation, as shown in **Fig. 1**. We used a convex array probe with four elements, where the radius of the convex surface is 50 mm and the element width is 5 mm. The center frequency of the pulse is 2.0 MHz, the pulse length is 10λ at center frequency, and the average length T_{ave} = 20 μs. Scatterers are randomly located, and the density of scatterers is 5 units per mm². Red blood cells in the vessel move forward to the probe and have a constant velocity of 0.5 m/s. We assume that scattering intensity of tissue scatterers is 30 dB higher than the moving scatterers in the vessel, and that the intensity of time varying components is 40 dB lower than intensity of all interferences and a cranium returns high-intensity plane wave.

3. Results

Fig. 2 shows the blood flow velocity for 10 samples estimated using the conventional method and the proposed method, where the cranium interference arrived from an incident angle of 6 degrees. To estimate blood flow velocity we use three pulses which is the minimum sample number for blood flow velocity estimation with MTI filter. When the desired signal intensity from red blood cells normalized by the signal intensity from the cranium interference (SI_CR) is less than –40 dB, the conventional method fails to estimate. Contrary, our proposed method succeeded in estimating a blood flow velocity of 0.5 m/s, with an average estimation error of less than 0.2 m/s, even when the SI_CR is –60 dB.

Fig. 3 shows the beam pattern calculated by the conventional method and the proposed method, where the SI_CR is –50 dB and the incident angle of the cranium interference is 10 degree. Our proposed method succeeds to form a null at the angle of the cranium interference.
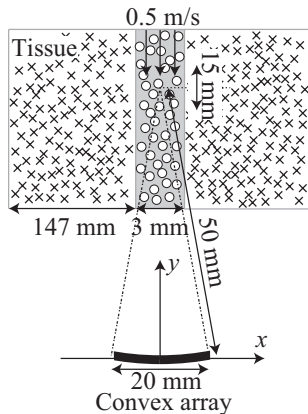
4. Conclusion

These results indicate the potential of the proposed method to improve the accuracy of blood flow velocity estimation under the condition that a high-intensity interference exists. We believe that the proposed method is effective for TCD.

References
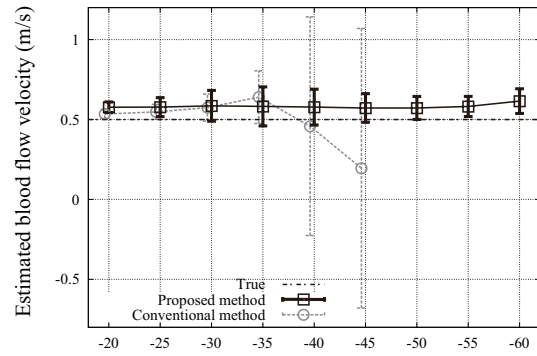1. S. Okumura, A. Kita, H. Taki, T. Sato: Proc. IEEE Ultrasonics Symp., 2013 accepted.

Fig. 2. Estimated blood flow velocities using a conventional method and the proposed method, where the signal intensity from blood cells normalized by the intensity of the cranium interference ranges from –20 to –60 dB. The true blood flow velocity is 0.5 m/s. Error bars show the standard deviation.
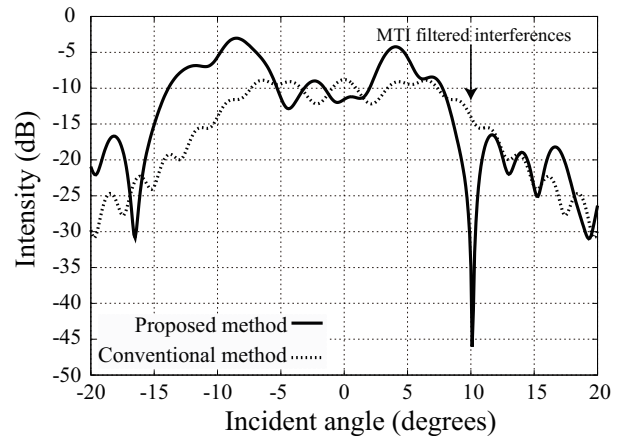

Fig. 3. Beam pattern calculated using the proposed method and the conventional method. The interference arrives from an incident angle of 10 degrees.

Fig. 1. Schematic illustration of system model.