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A Method of Automatically Detecting of Intima-Media Complex for Ultrasonic Measurement of Carotid Luminal Surface Roughness

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1. Introduction

For early detection of atherosclerosis, we have been developing a method to measure the minute surface roughness of intima-media complex (IMC) using the longitudinal displacement of IMC naturally caused by pulsation.[1] In our previous study, it was necessary to manually set the threshold for detecting IMC. However, automatic detection is desirable for clinical applications. In the present study, we investigated an automatic IMC detecting method for surface roughness measurement.

2. Methods

2.1 Ultrasonic measurement of carotid luminal surface roughness.[1]

The minute surface roughness is measured using a fixed ultrasound beam by utilizing the naturally occurring longitudinal displacement of IMC caused by pulsation. This method can measure the surface roughness for each beam without being affected by the region of non-uniform sound speed by combining the naturally occurring longitudinal displacement caused by pulsation and the axial displacement caused by the difference in the surface roughness in the longitudinal direction. The longitudinal and axial displacements are measured by the block-matching method[2] and the phased-tracking method[3], respectively. The longitudinal displacement of IMC during one heartbeat is larger than the beam spacing (0.15 mm in this study). Therefore, the measurement region of surface roughness can be increased by combining it measured for each ultrasound beam.

2.2 Automated detection of IMC

At the initial frame in the surface roughness measurement, the envelope signals reflected at (x_m, z_i) in the carotid artery are used. Here, x_m is the lateral position of the m-th ultrasonic beam and z_i is the depth of the i-th sample point, and the total numbers of beams and sample points are M and N, respectively. Then, these signals are normalized by the maximum amplitude in all M beams and all N sampled depths to obtain the signal $s(z_i; x_m)$. **Figure 1(a)** shows the B-mode image of the human carotid artery and **Fig. 1(b)** shows the normalized envelope signal $s(z_i; x_{20})$ at the beam position x_{20}. The reflected signals from IMC are normally separated from those from adventitia. Therefore, by searching from the lumen depth z_{lumen} to the deeper position, the depth $\hat{z}(x_m; T_\text{h})$ at which the amplitude of the normalized envelope signal $s(z_i; x_m)$ exceeds the threshold T_h for the first time and becomes local maximum is defined as the IMC depth, as shown in Eq. (1).

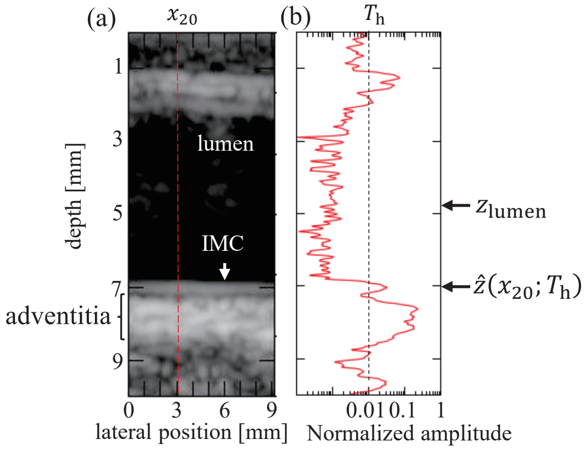

Fig. 1 (a) B-mode image of the human carotid artery, and (b) the normalized envelope signal $\boldsymbol{s(z_i; x_{20})}$ at beam position $\boldsymbol{x_{20}}$.

$$\hat{z}(x_m; T_\text{h}) = \min_{z_i > z_{\text{lumen}}} z_{\text{detect}},$$
$$z_{\text{detect}} = \{z_i | s(z_{i-1}; x_m) < s(z_i; x_m) \cap s(z_{i+1}; x_m) < s(z_i; x_m) \cap s(z_i; x_m) > T_\text{h}\}. \quad (1)$$

However, if the threshold T_h is not appropriate, the detected depth $\hat{z}(x_m; T_\text{h})$ is set in the lumen or adventitia. In this case, the detected depth becomes discontinuous in the lateral direction compared to when the IMC is correctly detected. Therefore, the discontinuity of detected depth at each threshold T_h is evaluated by summing the absolute differences of detected depth obtained for each adjacent beam for all beams, as shown in Eq. (2).

$$D(T_\text{h}) = \sum_{m=1}^{M-1} |\hat{z}(x_m; T_h) - \hat{z}(x_{m-1}; T_h)|. \quad (2)$$

The depth of IMC $\hat{z}_{\text{IMC}}(x_m)$ is determined by the optimal threshold $\widehat{T_\text{h}}$ at which the value of the discontinuity evaluation function $D(T_h)$ takes minimum as shown in Eq. (3).

$$\hat{z}_{\text{IMC}}(x_m) = \hat{z}(x_m; \widehat{T_h}),$$
$$\widehat{T_h} = \arg\min D(T_h). \quad (3)$$

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3. Experiment

The proposed method described in Sec. 2.2 was applied to the right common carotid artery of a healthy male in his 20s to detect IMC automatically. The luminal surface roughness on the detected IMC was measured by the method described in Sec. 2.1. An ultrasound diagnostic apparatus (ProSound F75; Hitachi Aloka) with a linear array probe (UST-5415; Hitachi Aloka) was used for the ultrasonic measurements. The transmitted ultrasound frequency and sampling frequency of the received signals were 7.5 and 40 MHz, respectively. The longitudinal cross-section of the IMC was measured with 61 ultrasonic beams. The beam spacing was 0.15 mm, and the frame rate was 187 Hz.

4. Result

Figure 2(a) shows the value of discontinuity evaluation function (purple line) of the detected depth for each threshold and their averaged depth for all beams (dashed line). The value of discontinuity evaluation function was minimized when the threshold was approximately 0.01 to 0.02. Since the detected depth $\hat{z}(x_m; T_\mathrm{h})$ calculated by Eq. (1) was the same for each beam from the threshold of 0.01 to 0.02, the discontinuity evaluation function $D(T_\mathrm{h})$ took the same value. When the threshold was less than approximately 0.01 or greater than approximately 0.02, the discontinuity evaluation function became larger, and the average detected depth for all beams became shallower and deeper, respectively. **Figure 2(b)** shows the detected depth of IMC (red line) using the optimized threshold $\widehat{T_\mathrm{h}}$, when the value of discontinuity evaluation function $D(T_\mathrm{h})$ was minimum. The detected depth of IMC was overlaid on the B-mode image of the posterior wall of the carotid artery. The detected depth could be set on the correct IMC for all beams. **Figure 2(c)** shows the surface roughness measured on the IMC by the method described in Sec. 2.1. The depth of IMC was detected automatically by the proposed method described in Sec. 2.2, and then for the determined surface, the roughness of the carotid luminal surface with the micron order was measured.

5. Discussion

Figure 3 shows the detected depth (red line) for $T_\mathrm{h} = 0.025$, when the value of discontinuity evaluation function was not minimum. **Figure 4** shows the longitudinal displacement of IMC during one heartbeat measured by the block-matching method[2]. As shown in Fig. 3, if the peak of the envelope signal in the IMC is smaller than the threshold, the proposed method cannot correctly detect the IMC. However, the areas where IMC could not be detected were within four consecutive beams (0.6 mm). As shown in Fig. 4, the IMC was displaced in the longitudinal direction by a maximum of 0.6 mm during one heartbeat in this

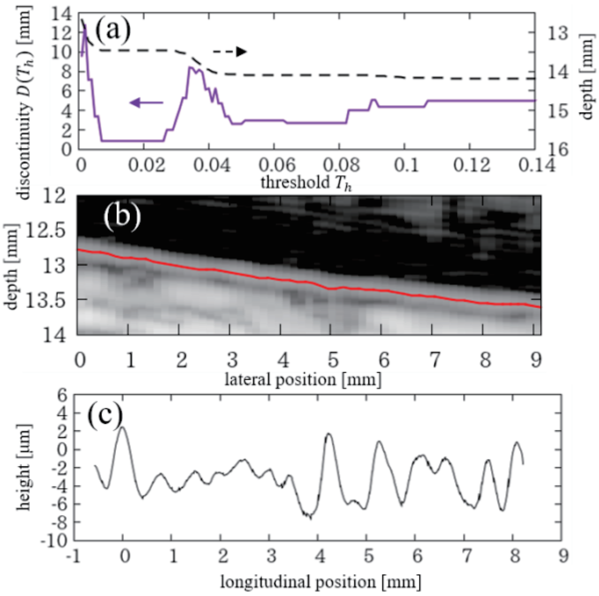

Fig. 2 (a) The value of discontinuity evaluation function $\boldsymbol{D(T_\mathrm{h})}$ (purple line) and their averaged depth for all beams (dashed line), (b) the detected depth of IMC (red line) using the threshold $\widehat{\boldsymbol{T_\mathrm{h}}}$, and (c) the measured surface roughness.

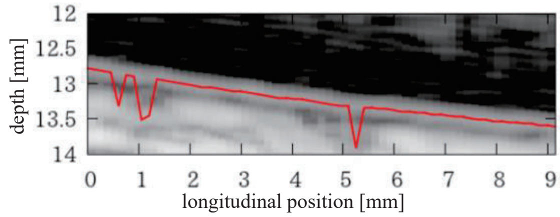

Fig. 3 The detected depth (red line) when threshold $\boldsymbol{T_\mathrm{h} = 0.025}$.

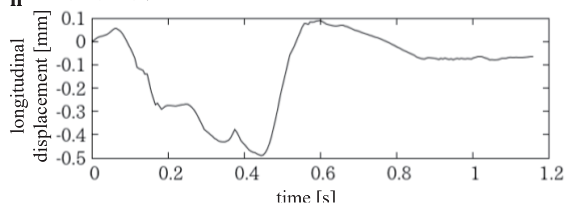

Fig. 4 The longitudinal displacement of IMC during one heartbeat.

subject. In this case, the surface roughness should be obtained by combining only the results with the beams correctly detecting the IMC.

6. Conclusion

The depth of IMC was automatically detected by the proposed method. This method contributes to the automation of the surface roughness measurement system.

References

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