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Detection of Arterial Wall Boundaries Using an Echo Model Composed of Multiple Ultrasonic Pulses

複数超音波パルスから構成された反射波モデルを用いた頚動脈壁の境界検出

Nabilah Ibrahim[1 ‡], Hideyuki Hasegawa[2,1] and Hiroshi Kanai[1,2]
([1]Grad. School of Eng.; Tohoku Univ. ,[2]Grad. School of Biomed. Eng.; Tohoku Univ.)
ナビラ　イブラヒム [1 ‡]，長谷川英之 [2,1]，金井　浩 [1,2] ([1] 東北大院　工; [2] 東北大院　医工)

1. Introduction

The atherosclerosis is considered to be the main cause of the cardiovascular diseases and the increase of the intima-media thickness (IMT) of the carotid artery is one of the early symptoms of the atherosclerosis lesion.[1)] Moreover, IMT correlates very well with pathohistologic measurements for assessing cardiovascular risk,[2)] where the IMT is widely defined by the distance between the lumen-intima boundary (LIB) and the media-adventitia boundary (MAB) of the posterior carotid artery wall. Thus, the IMT measurement is essential for the early diagnose of atherosclerosis. To overcome the drawback of manual method,[3)] which is lack of reproducibility, previous efforts have been done to automatically assess the LIB and MAB positions using template matching between the echo models (LIB and MAB echo models) and the measured *in vivo* RF signal.[4,5)] The positions of the models, which give the minimum difference between the measured and model signals, are determined as the boundaries. However, the approach detects the boundaries positions accurately in the case when the echo from the arterial wall is composed only of the echoes from LIB and MAB. In fact, echoes from the intima-media complex (IMC) and that from the scatterers below adventitia layer are also exist and, therefore, there is a limitation.

In the present study, the reference RF echo[7)] was used to create the optimum echo model. The echo model is improved from the previous one[4-6)] by creating the reference RF echo from the transmitted ultrasonic pulse measured with a hydrophone. Since the redundant echo is present in the measured echo from arterial wall, multiple reference RF echoes (ultrasonic pulses) are applied to compose the echo model. The procedure to perform the detection of the carotid arterial wall boundaries using the technique proposed by our group is described in the subsequent sections.

2. Simulation of the Echo Model by Imitating the Reference RF Echo

The measurement of the reference RF echo which is calculated from the transmitted ultrasonic pulse measured with a hydrophone has been reported previously.[7)] By imitating the reference RF echo, the echo model is created. The echo model is created by multiplying the sinusoidal wave to the envelope of the reference RF echo (reference envelope). To create the optimum echo model, the optimum center frequency needs to be determined. For the estimation of the center frequency, the initial phase of the sinusoidal wave is changed within the reference envelope to find the frequency which gives the maximum correlation coefficient between the echo model and the reference RF echo. From the distribution of the correlation coefficient between the reference RF echo and the echo model in Fig. 1, at the maximum correlation coefficient 0.995, the frequency of 7.4 MHz is determined. Figure 2(a) shows the reference RF echo and created echo model at the maximum correlation. Thereby, the echo model that is necessary to be fitted with the measured *in vivo* RF signal consists of 5 cycles as shown in Fig. 2(b), where the leading edge of the echo model $\hat{s}(t)$ is determined at nearly zero phase of the model.

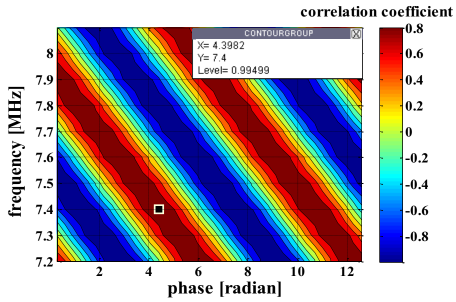

Fig. 1 Distribution of correlation coefficient between the reference RF echo and echo model.

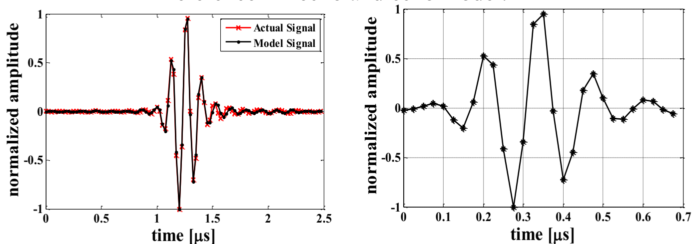

Fig. 2 (a) Actual and model signals at the maximum correlation (0.995). (b) Simulated echo model.

3. Procedure of Fitting Echo Models to Echo Signals from Arterial Wall

Our proposed automated template matching method for the detection of the carotid arterial wall boundaries is realized by fitting the model signal $\hat{x}(nT_s)$ to the measured *in vivo* RF signal $x(nT_s)$ in eq. (1).

$$\alpha = \frac{1}{N}\sum_{m=0}^{N-1}\frac{\sum_{nT_s=T_m}^{T_{m+1}}\left|z(nT_s)-\hat{z}(nT_s)\right|^2}{\sum_{nT_s=T_m}^{T_{m+1}}\left|z(nT_s)\right|^2}+\frac{\sum_{nT_s\in L}\left|x(nT_s)-\bar{x}\right|^2}{\sigma_i^2}$$
$$+\sum_{nT_s=T_0}^{T_1}\frac{\left|x(nT_s)-\bar{x}\right|^2}{\sigma_j^2}, \qquad (1)$$

where $\hat{z}(nT_s)$ and $z(nT_s)$ are complex signals of $\hat{x}(nT_s)$ and $x(nT_s)$, respectively, which are obtained using the

E-mail: nabilah@us.ecei.tohoku.ac.jp
{hasegawa, kanai}@ecei.tohoku.ac.jp

Hilbert transform. Since the complex signals are employed, the phase information of the echo from the LIB and that from the MAB are considered. The normalized MSE α is estimated and normalized with respect to $\{T_m\}$ as shown in Fig. 3. Since the redundant echo from the IMC and the number of scatterers that present below the adventitia layer are unknown, the model signal that composed of multiple ultrasonic pulses should be employed instead of that is composed of two pulses (LIB and MAB echo models). However, it is reported that the normal IMT is within the range between 0.5 and 1.0 mm[8] and is varied from person to person.[9] Hence, in this work, the fitting region was scaled between 0.3 mm and 1.0 mm. Thus, the maximum number of pulses to construct the model signal $\hat{x}(nT_s)$is four [eq. (2)]. However, the number of pulses is determined so that the minimum normalized MSE α is minimized.

$$\hat{x}(nT_s) = \sum_{m=1}^{k} \hat{s}(nT_s - \tau_m), \qquad (k = 1,2,3,4). \qquad (2)$$

Additionally, the variance between the model signals in case when the model signals do not overlap is calculated and normalized by the variance σ_i^2 of the measured RF signal in the whole analyzed region as shown by the second term on the right hand-side of eq. (1). The normalized variance increases when the model signals are fitted with the echo from the tissue below the adventitia layer that is false detection. Therefore, the estimation of the normalized variance could prevent the echo from the below adventitia layer from being detected. For the robust detection of LIB, the variance β in the region above the LIB echo is also added (third term in eq. (1)). The variance β is normalized by the variance σ_j^2 in the region of interest before the LIB echo model. Since the magnitude of the echoes from the lumen is small, β is nearly zero. On the other hand, β increases when LIB echo model is fitted to the MAB echo. Therefore, the misdetection of the LIB and MAB can be reduced by calculating such the variances.

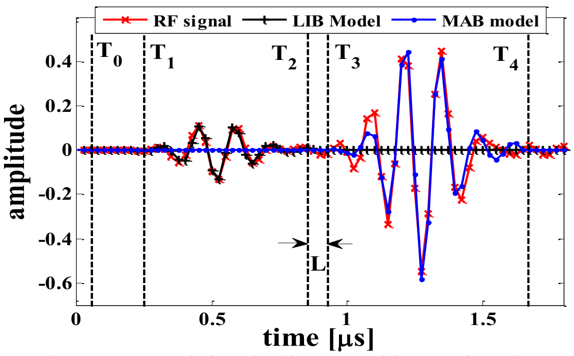

Fig. 3 Measured signal of beam position *i* [Fig. 4(b)] and echo model composed of two pulses at minimum MSE α.

Thereafter, the positions of the models which give the minimum difference between the measured and model signals are determined as the optimum position of the LIB and MAB. In detail, the position of the MAB is determined at the position of the fitted model signal with maximum amplitude, in the region of IMC and adventitia layer. The determination was performed under the consideration that the adventitia results in reflection with a high-amplitude signal.

4. Results and Conclusion

Figure 4(a) shows the B-mode image with detected boundaries of the carotid arterial wall when the echo model is composed of two pulses in the previous method.[7] Figure 4(b) shows the results of detected boundaries by the new approach of the proposed method.

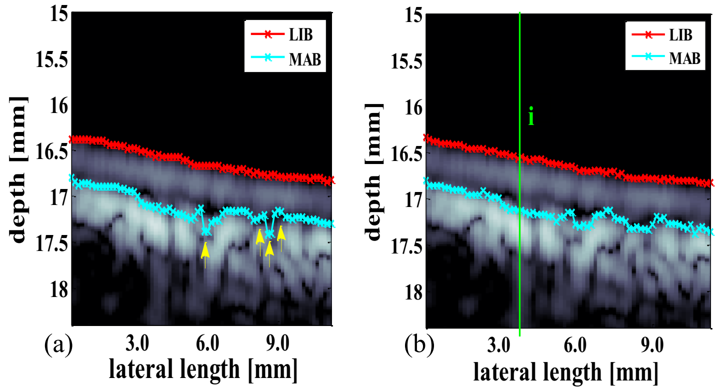

Fig. 4 Detected boundaries of posterior carotid arterial wall for same subject by (a) previous method[7] and (b) proposed method.

From the above results, at some beam positions (marked by arrows), the estimated boundaries by the new approach of the proposed method differ from those used two model signals. For those beams, the percent minimum α is significantly reduced from [18.1% - 35.7%] to [2.2% - 13.0%]. For all analyzed beams, the only [1.2% - 13.0%] of α convinces that the model signal composed of multiple pulses is very good agreement with the measured *in vivo* RF signal. Thus, the proposed method leads to the accurate boundaries detection. The average IMT for 11.0-mm-long short segment in the arterial longitudinal direction is 516±52 μm. To evaluate the reliability of the proposed method, the *in vivo* experiment and the boundaries detection of IMC for other subjects are going to be conducted.

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