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# Improvement of Accuracy in Ultrasonic Measurement of Luminal Surface Roughness of Carotid Arterial Wall by Deconvolution Filtering

## 頸動脈内膜表面粗さの超音波計測におけるフィルタ処理による高精度化

Yoshifumi Nagai[1‡], Hideyuki Hasegawa[1, 2] and Hiroshi Kanai[2, 1] ([1] Grad. School of Biomed. Eng., Tohoku Univ.; [2] Grad. School of Eng., Tohoku Univ.)
永井慶史 [1‡], 長谷川英之 [1, 2], 金井　浩 [2, 1] ([1] 東北大院　医工; [2] 東北大院　工)

### 1. Introduction

It is reported that the internal elastic layer and endothelial cells are damaged in the early stage of atherosclerosis[1)]. As a result, the luminal surface becomes rough. Therefore, it would be useful to measure such minute roughness on luminal surface of the arterial wall for diagnosis of early-stage atherosclerosis. The luminal surface on the arterial wall is covered with endothelial cells, and the thickness of an endothelial cell is 10-20 μm in thickness[2)]. Therefore, high spatial resolution of micron order is required to measure such surface roughness. In the present study, we proposed a method to measure minute surface roughness accurately by ultrasound with filtering.

### 2. Principle

**Figure 1** illustrates displacements of the arterial wall caused by heartbeat, where the axial and lateral directions are shown by $z$-axis and $x$-axis, respectively. The lateral position of the $m$-th ultrasonic beam is denoted by $x_m$. During a cardiac cycle, the arterial diameter is expanded due to an increase of internal pressure at the arrival of the pulse wave and the arterial wall is displaced in the longitudinal direction along $z$-axis[3)] because the arch of aorta is pulled due to contraction of the heart. The axial displacement of the surface of the posterior wall $\Delta d(x_m, n) \equiv \Delta d(m,n)$ between the $n$-th frame ($t$ [s]) and ($n$+1)-th frame (($t$+$\Delta T$) [s]) is obtained from the difference between the axial position $z(x_m, n) \equiv \Delta z(m,n)$ of the surface in the $n$-th frame and that $z(x_m, n+1) \equiv \Delta z(m, n+1)$ in the ($n$+1)-th frame. The axial displacement $d(m,n)$ between the first frame and the $n$-th frame is calculated by accumulating instantaneous displacements $\{\Delta\hat{d}(m,n)\}$ estimated by the

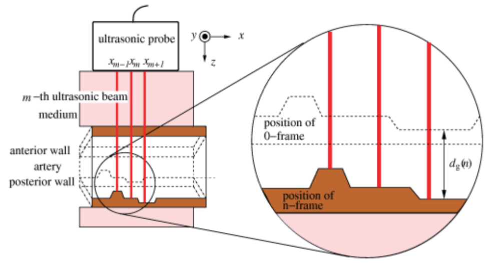


Fig. 1. Illustration of principle of measurement.

*phased-tracking method*[4)].

The measured instantaneous displacement $\Delta\hat{d}(m,n)$ contains both the global displacement $\Delta\hat{d}_g(n)$ of the arterial wall due to expansion of the artery and the displacement $\Delta\hat{d}_s(m,n)$ due to the surface roughness. Compared with the wavelength of the pulse wave (tens of centimeters), the length of a measured region (a few millimeters) is minute. Therefore, the axial displacements $\{d_g(m+i, n)\}$ ($i$ = 0, 1, 2, …, $M$-1) caused by global wall motion between the first and $n$-th frames at beam positions $\{x_{m+i}\}$ are assumed to be same. The global displacement $d_g(n)$ can be estimated by averaging the axial displacements $\{d_g(m+i, n)\}$ ($i$ = 0, 1, 2, …, $M$-1) at all beam positions. Therefore, the displacement $d_s(m, n)$ caused by the surface profile can be estimated by subtracting the global displacement $d_g(n)$ from measured displacement $\hat{d}(m,n)$ as follows:

$$\hat{d}_s(m,n) = \hat{d}(m,n) - \hat{d}_g(n), \quad (1)$$

The surface profile $z_m(x; z_{m0})$ estimated at the $m$-th ultrasonic beam is expressed as follows:

$$\begin{aligned}\hat{z}_m(x; z_{m0}) &\equiv \hat{z}_m(x_m + \hat{l}(m,n); z_{m0}) \\ &= z_{m0} + \hat{d}_s(m,n),\end{aligned} \quad (2)$$

where $z_{m0}$ is an initial axial height.

Using the block matching, the lateral displacement $l(m,n)$ of the arterial wall at the $m$-th ultrasonic beam between the first and $n$-th frames is estimated. Using the estimated lateral

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E-mail address: nagai@us.ecei.tohoku.ac.jp
(kanai, hasegawa)@ecei.tohoku.ac.jp

displacement $\hat{l}(m,n)$, the lateral position of a point on the arterial wall where the $m$-th ultrasonic beam crosses in the $n$-th frame is determined.

The arterial movement is small (1 mm) and, thus, the measured region is limited. Therefore, surface roughness was measured in multiple ultrasonic beams, and the measured surface profiles were combined to obtain the surface profile over a wider region[4)].

The surface profile measured by ultrasound degraded by the point spread function (PSF) of the ultrasonic diagnostic system. We developed a filter $M(\omega)$, which is based on the Wiener filter, to eliminate PSF. **Figure 2** shows a block diagram illustrating how to estimate the true surface profile. An echo from a point scatterer of fine wire of 15 µm in diameter was used to determine PSF $H(\omega)$ and inverse filter for deconvolution was designed as follows:

$$M(\omega)=\frac{1}{H(\omega)}\frac{|H(\omega)|^2}{|H(\omega)|^2+\beta(P_N(\omega)/P_F(\omega))}, \qquad (3)$$

where $P_N(\omega)$ and $P_F(\omega)$ are power spectra of noise $N(\omega)$ and true signal $F(\omega)$, and $\beta$ is the control parameter. The estimate $\hat{F}(\omega)$ is given by

$$\hat{F}(\omega)=G(\omega)\cdot M(\omega)\ , \qquad (4)$$

where $G(\omega)$ is the spectra of the original signal obtained by measurements. By inverse Fourier transformation, surface profile $\hat{f}(\omega)$ is estimated.

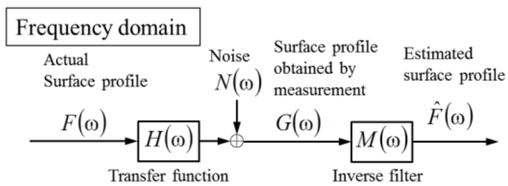


Fig. 2. Block diagram of the system for obtaining estimate $\hat{F}(\omega)$ using inverse filter $M(\omega)$.

## 3. Results

In experiments, a silicone phantom which has minute saw-teeth shapes (height of a tooth is about 12 µm) was measured. **Figure 3** shows a B-mode image of the surface of the phantom (surface profile in red region is shown in **Fig. 4**). **Figure 4(a)** shows the true surface profile measured by a laser profilometer and **Fig. 4(b)** shows the original measured surface profile (before deconvolution, blue line) and the estimated surface profile (after deconvolution, pink line). This result showed that the surface roughness of the phantom could be estimated with better spatial resolution using filtering. The surface profile, which loses its minute serrated shape, were sharpened and improved by the proposed filtering. The deconvolution filter is applied to data measured *in vivo*. **Figure 5** shows the B-mode image and the luminal surfaces profile of carotid arterial wall. The data was acquired from a healthy subject (23 years old) in *in vivo* experiments.

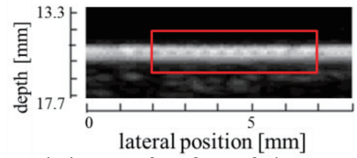


Fig. 3. B-mode image of surface of phantom. Estimated surface profile in red region is shown in Fig. 4.

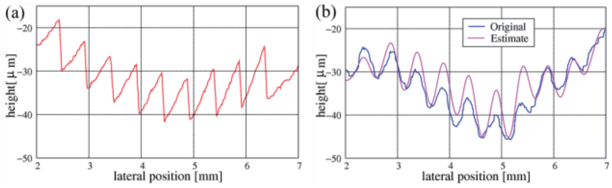


Fig. 4. (a) True surface profile $f(x)$ measured by a laser profilometer. (b) Original surface profile $g(x)$ (before deconvolution, blue line) and estimated surface profile $\hat{f}(x)$ (after deconvolution, pink line).

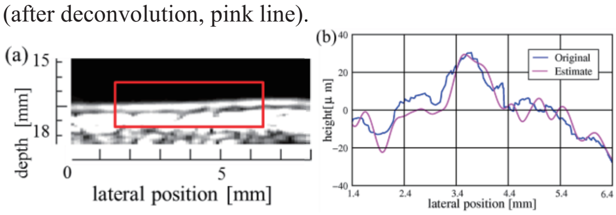


Fig. 5. (a) B-mode image of luminal surface. Estimated surface profile in red region is shown in Fig. 5(b). (b)Luminal surface profile of the carotid arterial wall. Blue line is the original surface profile $g(x)$ (before deconvolution) and pink line is the estimated surface profile $\hat{f}(x)$ (after deconvolution).

## 4. Conclusion

In this study, the phantom which has minute surface roughness is measured to evaluate the effect of the proposed deconvolution filter. From the measured PSF, an inverse filter for deconvolution was designed. By using the proposed filter, the surface profile, which loses its minute serrated shape, was improved.

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