J Med Ultrasonics (2007) 34:23–27
DOI 10.1007/s10396-006-0125-8

ORIGINAL ARTICLE

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Improved imaging of the carotid artery in the short-axis plane by a mechanical scanning ultrasonic probe

Received: June 2, 2006 / Accepted: August 7, 2006

Abstract To image the intima–media complex of the carotid artery in a wider region, a method for measuring cross-sectional images in the arterial short-axis plane is presented. Using the proposed mechanical scanning system for an ultrasonic probe, cross-sectional images of a silicon rubber tube and a human carotid artery are measured in basic experiments and in in vivo experiments, respectively. These experiments show that this method successfully images the short-axis cross sections. Using the method proposed in this article, B-mode images in the short-axis plane can be accurately measured in a wider region than is possible with conventional methods.

Keywords B-mode image of artery · short-axis plane · intima–media thickness · ultrasound

Introduction

Recently, the increase in the number of patients suffering from cardiovascular or cerebrovascular diseases has become a serious social problem. Because such circulatory diseases are mainly caused by atherosclerosis, it is important to diagnose this disease in an early stage. Intima–media thickness (IMT) of the carotid artery noninvasively measured with conventional ultrasonic diagnostic equipment is a useful marker for diagnosis of atherosclerosis.[1–3] Bots et al. showed that IMT increases inhomogeneously, developing atherosclerosis plaque.[1] Therefore, the intima–media complex should be imaged omnidirectionally. However, when an artery is imaged in the short-axis plane, which is the plane perpendicular to the axis of the artery, with conventional linear scanning, ultrasonic beams that do not pass through the center of the artery are not perpendicular to the arterial wall. To image an object clearly, it is important that ultrasonic beams are perpendicular to an object. Wilhjelm et al. measured the power of the received echo, $E(\theta)$, while varying the angle of incidence, θ.[4] In their report, in the case of a smooth interface placed at the focal point, the maximum power was measured at $\theta = 90°$, and the width at half-maximum was determined to be $\pm 4°$ using a 5-MHz concave transducer at a 25.4-mm diameter. From these facts, IMT cannot be measured accurately when ultrasonic beams are not perpendicular to the wall because the decrease in the echo amplitude makes the B-mode image unclear. Therefore, it is very important that ultrasonic beams are perpendicular to the wall for measurement of IMT.

The cross-sectional image of the arterial wall in the short-axis plane is measured by intravascular ultrasonography (IVUS). Nissen et al. reported that the size and shape of the atherosclerotic plaque on the arterial wall could be observed in B-mode images acquired by IVUS.[5] de Korte et al. developed a method using IVUS for measuring the strain of the arterial wall caused by the change in inner pressure.[6] Furthermore, Mita et al. measured the strain during an entire heart cycle by compensating for the change in the positions of the IVUS probe.[7] However, IVUS cannot be applied to noninvasive measurement of the carotid artery, which is the most suitable region for measurement of IMT.[3] Therefore, a method for noninvasive measurement of IMT in the short-axis plane should be developed. To make all the ultrasonic beams perpendicular to the wall, Nakagawa et al. proposed a new scanning method in which ultrasonic beams are applied to a plane perpendicular to the axis of the artery (short-axis plane).[8] Although this method extends the region that can be imaged, the region is still limited to 45° because the increase in the grating lobe leads to a loss of precision.

In this study, we constructed a system with a mechanical scanning ultrasonic probe to image the intima–media complex in a wider region in combination with the method proposed by Nakagawa et al.[8] Using this mechanical scanning system, we measured cross-sectional (B-mode) images of a silicon rubber tube in basic experiments and a human common carotid artery in in vivo experiments. The results

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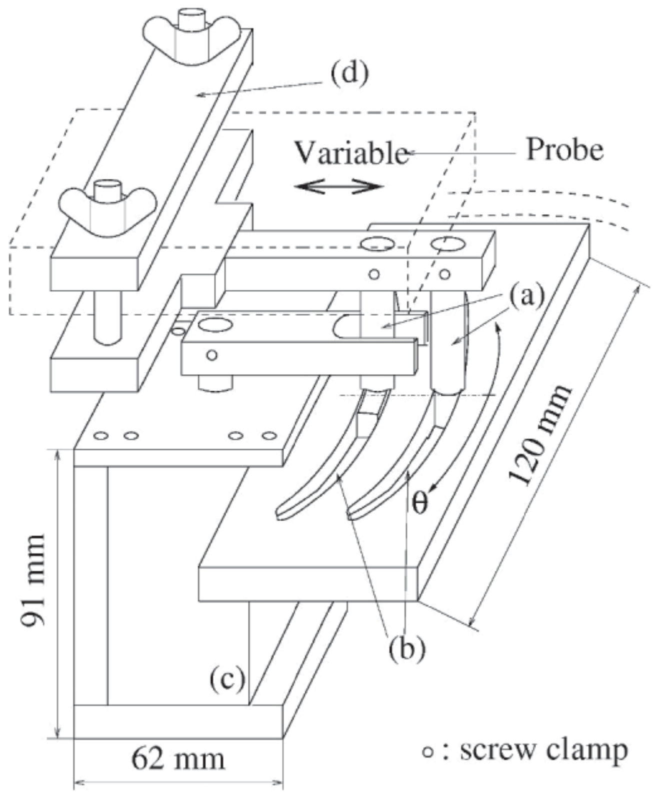

Fig. 1. Schematic diagram of the mechanical probe shows round bars (*a*), rails (*b*), stepping motor (*c*), and device to hold a linear-type probe (*d*)

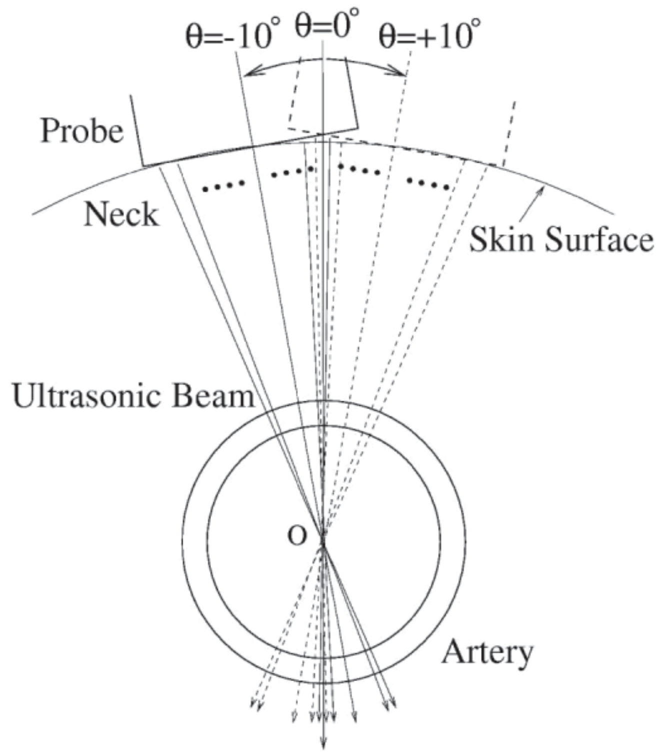

Fig. 2. Schematic diagram of mechanical scanning with an ultrasonic probe

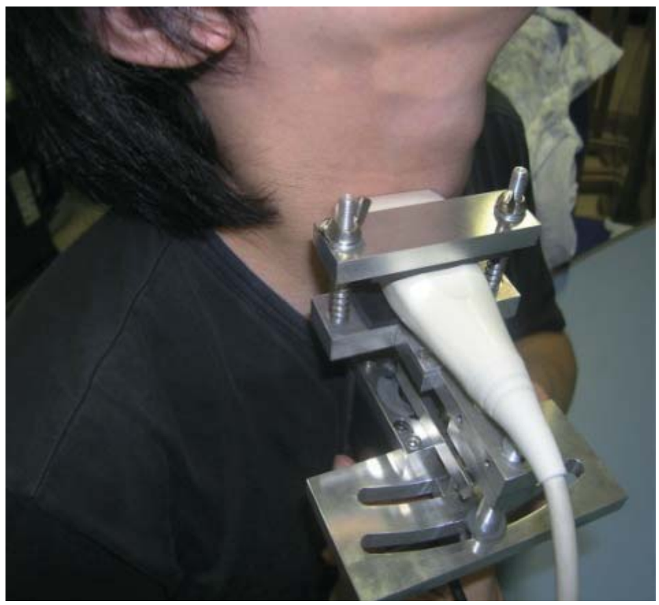

Fig. 3. Appearance of in vivo experiment

showed that the intima–media complex can be imaged in a wider region in the short-axis plane.

Methods

Experimental system

Figure 1 shows a schematic diagram of the mechanical scanning system, which mainly consists of round bars, rails, a stepping motor, and a device to hold a linear-type ultrasonic probe. By rotating the stepping motor (RK544AA-N10; Oriental Motor, Tokyo, Japan), the round bars (Fig. 1, *a*) move along the rails (Fig. 1, *b*). These rails are in the form of concentric circular arcs. In preliminary measurements, we measured the carotid artery in four healthy men. As the distance from the ultrasonic probe to the center of the artery was 9.0–13.1 mm in these subjects, we designed this mechanical probe to enable measurement of the IMT when the distance from the ultrasonic probe to the center of the artery is 8.0–15.0 mm. As shown in Fig. 1, we can adjust the position where the probe is fixed to match the distance from the skin surface to the artery. This adjustment of the probe position enables the ultrasonic probe to move along a circular arc whose center coincides with the center of the artery. At each mechanical rotation angle, θ (Fig. 2), the ultrasonic beam electronically scans the region of ± 22.5° so as to be perpendicular to the arterial wall. Using this mechanical scanning system, ultrasonic beams become perpendicular to the wall in the region of 65° when the ultrasonic probe moves at the maximum angle. The device is small and light enough to be held in one hand (Fig. 3). Various ultrasonic probes can be fixed to the mechanical scanning system.

In basic experiments and in vivo experiments, a conventional linear-type ultrasonic probe was used (center

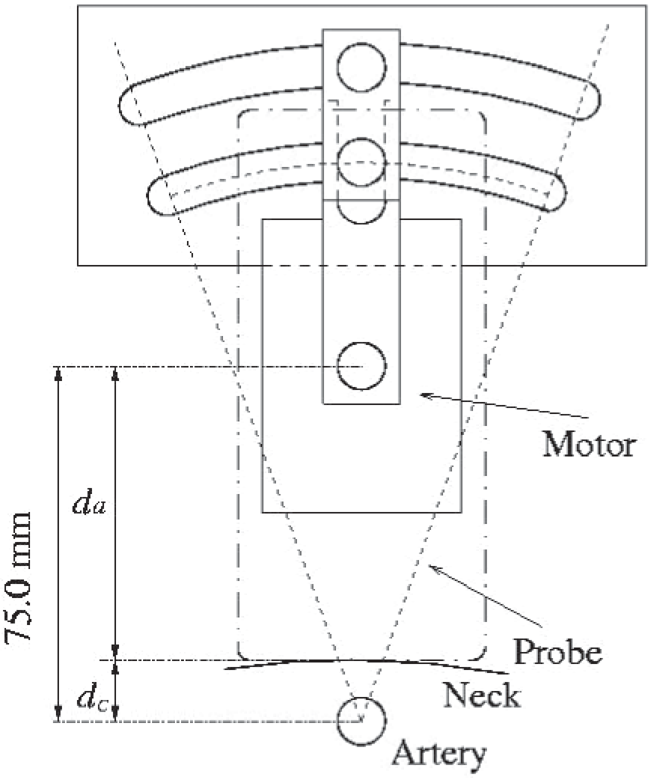

Fig. 4. Schematic diagram of mechanical scanning system

frequency, $f_0 = 10$ MHz; interval of centers of apertures for transmitted beams, $L = 0.4$ mm; element pitch, $d = 0.2$ mm). The number of electronically scanned beams is 30 per frame. Ultrasonic radiofrequency (RF) echoes are sampled at 40 MHz with 16-bit resolution.

RF data acquisition and image reconstruction

Figure 4 shows a schematic diagram of the mechanical scanning system. The position of an ultrasonic probe can be changed in the depth direction (vertical direction in Fig. 4) so that the center of rotation of the ultrasonic probe coincides with the center of the carotid artery. Using the depth of the carotid artery, d_c, the position, d_a, where the ultrasonic probe is fixed to the mechanical scanning system, becomes $75.0 - d_c$ (mm). Thus, it is necessary that the depth of the carotid artery be measured. Therefore, the depth of the carotid artery is measured beforehand by the conventional linear scanning method, and the ultrasonic probe is attached to the position where $d_a = 75.0 - d_c$ (mm).

After attaching the probe, RF echoes from an object are acquired at a rotation angle of $\theta_r = 0°$, using the method proposed by Nakagawa et al.[8] Then, the ultrasonic probe is moved at a speed of 10°/s for t s, resulting in the rotation angle of $\theta_r = 10\ t$. After moving the probe, reflected RF echoes from the object are acquired by the same method. From these procedures, two B-mode images are acquired before and after moving the ultrasonic probe. Figure 5 shows the regions of RF data acquisition; there is a region where two B-mode images overlap because the rotation angle of the ultrasonic probe, θ_r, is smaller than the maximum beam angle, θ_{max} (= 22.5°). To reconstruct a B-mode image using RF data obtained by a couple of acquisitions, RF echoes obtained by beams with smaller steering angles are used in the region where B-mode images overlap because a smaller steering angle achieves a smaller grating lobe. Therefore (see Fig. 5), a B-mode image is constructed by RF echoes acquired in the region between $-\theta_{max}°$ and $+\theta_r/2°$ before moving the ultrasonic probe and between $-\theta_r/2°$ and $+\theta_{max}°$ after moving the probe. In this article, the rotation angles of the ultrasonic probe, θ_r, in basic experiments and in vivo experiments are 20° and 10°, respectively.

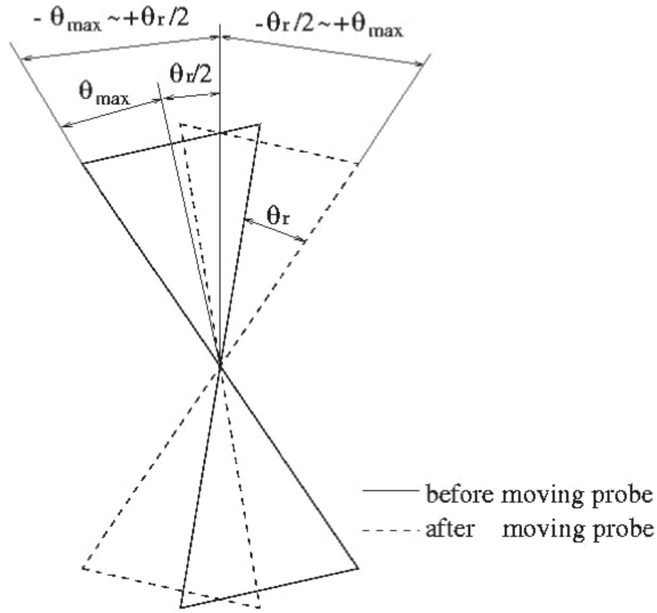

Fig. 5. Schematic diagram of the region of radiofrequency (RF) data acquisition

Results

Results of basic experiments using a silicon rubber tube

Figure 6a shows a B-mode image of a silicon rubber tube obtained by conventional linear scanning; echoes from the silicon rubber tube were detected only in the region of 30°. In this region, the angle of the beam to the wall changes by less than 30°/2 = 15° from the angle in the case of perpendicular insonification. In other regions, the wall was not detected.

Figure 6b shows a B-mode image of the same tube obtained by mechanical scanning with an ultrasonic probe, as proposed in Methods. The B-mode image shown in Fig. 6b was obtained by combining B-mode images at rotation angles of $\theta = -10°$ and 10°. Using the proposed method, echoes from the silicon rubber tube were clearly detected in the region of 65° because these ultrasonic beams become perpendicular to the wall of the silicon rubber tube. Therefore, using the scanning system proposed in this article, the region where the silicon rubber tube was imaged in the short-

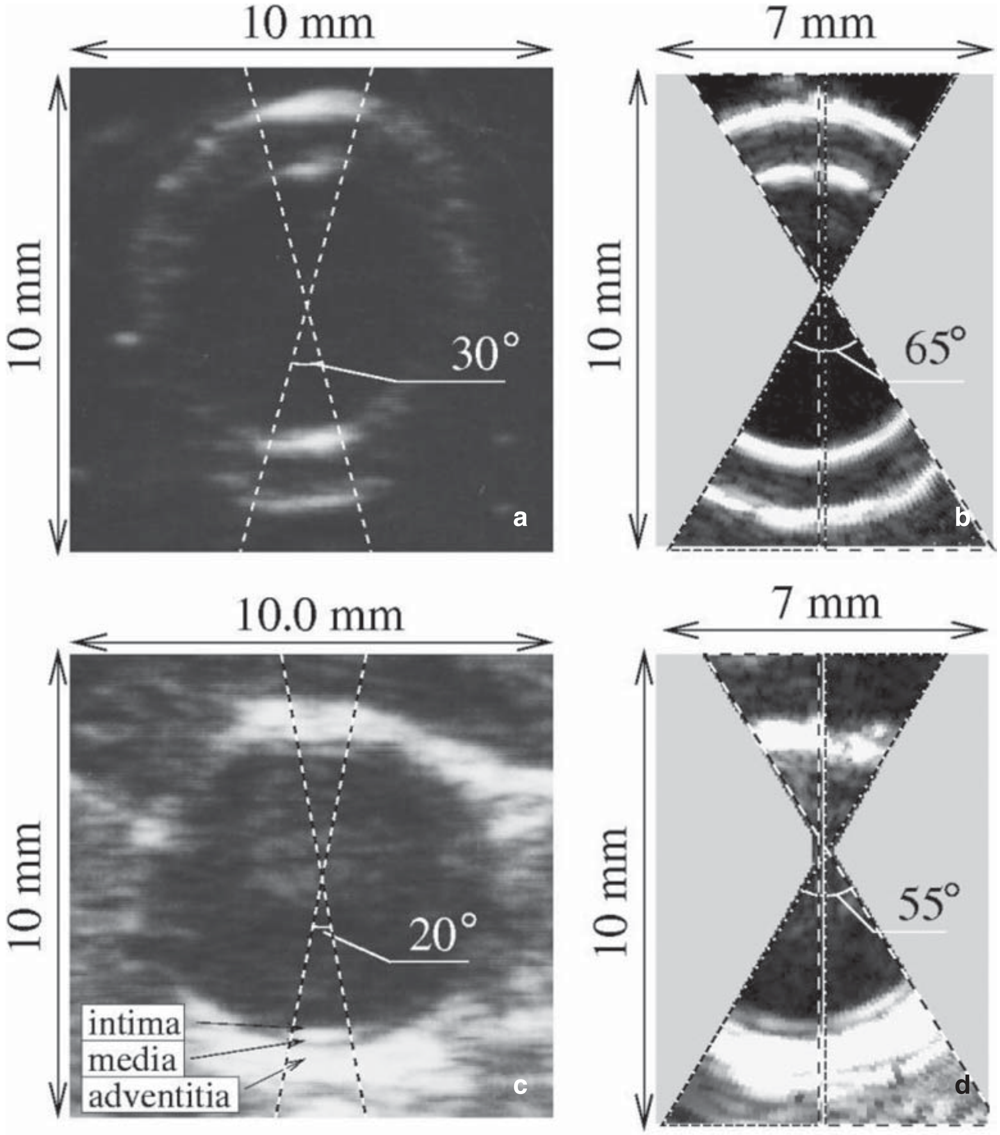

Fig. 6. B-mode images of a silicon rubber tube obtained by conventional linear scanning (**a**) and by the proposed method (**b**). B-mode images of a carotid artery of a 22-year-old healthy man were obtained by conventional linear scanning (**c**) and by the proposed method (**d**)

axis plane became 35° and 20° wider than those obtained by the conventional method and the method previously proposed by Nakagawa et al.,[8] respectively.

Results of in vivo measurements of a human carotid artery

Figure 6c shows a B-mode image of a human common carotid artery (CCA) of a 22-year-old healthy man obtained by conventional linear scanning. Using conventional linear scanning for in vivo experiments, the intima–media complex could not be measured in the region where the angle of ultrasonic beams changed by more than ± 10° (= 20°/2) from the angle in the case of perpendicular insonification to the arterial wall. In an in vivo experiment, the amplitudes of echoes from the lumen–intima interface were lower than those from the silicon rubber tube in basic experiments because there is not much difference in acoustic impedance between the blood and the intima at the interface. Therefore, the region where echoes were clearly detected in in vivo experiments was 10° narrower than the region in the basic experiments already presented.

Figure 6d shows a B-mode image of the same CCA obtained by mechanical scanning with an ultrasonic probe. This image was acquired by combining two B-mode images obtained at rotation angles of –5° and 5°. For in vivo measurement, the distance from the ultrasonic probe to the artery is slightly changed during mechanical scanning because the center of the neck does not coincide with that of an artery. Therefore, this change in distance was corrected based on the spatial cross-correlation between the RF echo lines along the axis shown by the white solid line in Fig. 6d, which were obtained at two different rotation angles, θ, of –5° and 5°. Using the mechanical probe, the intima–media complex can be accurately measured in a wider region, as was possible in the basic experiments. The IMT measured in this experiment was 0.6 mm, the same as with the conventional linear scanning method. As shown in Fig. 6d, the IMT can be measured in a region of 55°, which is 35° wider than that possible with the conventional linear scanning method, and it is 10° wider than that achieved by the method previously proposed by Nakagawa et al.[8]

Discussion

A method for steering the ultrasonic beam is used in this article. Thus, grating lobes are generated and may cause artifacts in a reconstructed image. Using the element pitch, d, width of element, a, wavelength, λ, and beam angle of main lobe, θ_0, the directional characteristic of an ultrasonic beam is defined as follows[9]:

$$R(\theta) = R_e(\theta) R_a(\theta) \tag{1}$$

where

$$R_e(\theta) = \frac{\sin\left(\frac{\pi a}{\lambda}\sin\theta\right)}{\frac{\pi a}{\lambda}\sin\theta}, \tag{2}$$

$$R_a(\theta) = \frac{\sin\left\{\frac{N\pi d}{\lambda}(\sin\theta - \sin\theta_0)\right\}}{N\sin\left\{\frac{\pi d}{\lambda}(\sin\theta - \sin\theta_0)\right\}}, \tag{3}$$

and N is the number of elements. From the above equations, an amplitude of a grating lobe is equivalent to that of the main lobe under the condition of $\sin\theta_0 = \lambda/2d$. In this study, the angle when an amplitude of the grating lobe is equivalent to that of the main lobe, θ_s, is ±22.5°, using the element pitch of $d = 0.2$ mm and wavelength of $\lambda = 150\,\mu$m. Therefore, in this study, the steering angle, θ_0, was restricted to be less than ±22.5°.

To suppress the grating lobe, the element pitch must be sufficiently narrower than the wavelength. However, for measurement of IMT, the spatial resolution of a B-mode image should be of submillimeter order.[1] Spatial resolution in the depth direction depends on the wavelength and is 75 (= 150/2) μm when the center frequency is 10 MHz. Thus, a larger wavelength is not desirable in terms of the spatial resolution. Therefore, reduction of the element pitch is a better solution for suppressing the grating lobe. Using current technology, the element pitch of a 10-MHz ultrasonic probe can be reduced to 0.125 mm.[10] In this case, θ_s becomes 36.9°. However, it is difficult to fabricate a probe with such a narrow element pitch, and it is not common at this moment. Therefore, the proposed mechanical scanning system is effective for imaging the carotid artery in the short-axis plane.

Table 1 shows the regions that can be accurately measured in the short-axis plane with the conventional linear scanning method, the method proposed by Nakagawa et al.,[8] and the method proposed here. Table 1 indicates that the method proposed in this article is an effective method for measuring IMT in a wider region in the short-axis plane.

Table 1. Comparison of regions scanned by conventional method, method proposed by Nakagawa et al.,[8] and method proposed in this article

Method	Scanned region (degrees)	
	Basic experiment	In vivo experiment
Conventional method	~30	~15
Method proposed by Nakagawa et al.	~45	~45
Method proposed in this article	~65	~55

Conclusions

In this study, we proposed a system for mechanically scanning with an ultrasonic probe to image the intima–media complex in a wider region in the short-axis plane in combination with the method previously proposed by Nakagawa et al.[8] The proposed method was evaluated by basic experiments using a silicon rubber tube and then applied to in vivo measurement of the human carotid artery. Using the proposed scanning method, B-mode images in the short-axis plane could be accurately measured in a wider region than is possible with the conventional linear scanning method and the previous method proposed by Nakagawa et al.

References

1. Bots ML, Hofman A, de Jong PTVM, et al. Common carotid intima-media thickness as an indicator of atherosclerosis at other sites of the carotid artery. Ann Epidemiol 1996;6(2):147–53.
2. van Swijndregt ADM, The SH, Gussenhoven EJ, et al. An in vitro evaluation of the line pattern of the near and far walls of carotid arteries using B-mode ultrasound. Ultrasound Med Biol 1996; 22(8):1007–15.
3. van Swijndregt ADM, de Lange EE, de Groot E, et al. An in vivo evaluation of the reproducibility of intima-media thickness measurements of the carotid artery segments using B-mode ultrasound. Ultrasound Med Biol 1999;25(3):323–30.
4. Wilhjelm JE, Pedersen PC, Jacobsen SM. The influence of roughness, angle, range, and transducer type on the echo signal from planar interfaces. IEEE Trans Ultrason Ferroelectr Freq Control 2001;48(2):511–21.
5. Nissen SE, Gurley JC, Grines CL, et al. Intravascular ultrasound assessment of lumen size and wall morphology in normal subjects and patients with coronary artery disease. Circulation 1991;84(3): 1087–99.
6. de Korte CL, Cespedes EI, van der Steen AFW, et al. Intravascular elasticity imaging using ultrasound: feasibility studies in phantoms. Ultrasound Med Biol 1997;23(5):735–46.
7. Mita H, Kanai H, Koiwa Y, et al. Cross-sectional elastic imaging of arterial wall using intravascular ultrasonography. Jpn J Appl Phys 2001;40(7):4753–62.
8. Nakagawa N, Hasegawa H, Kanai H. Cross-sectional elasticity imaging of carotid arterial wall in short-axis plane by transcutaneous ultrasound. Jpn J Appl Phys 2004;43(5B):3220–6.
9. Mahafza BR. Introduction to radar analysis. Huntsville, AL: Colsa Corporation; 1998. p. 270–5.
10. Dickinson RJ, Kitney RI. Miniature ultrasonic probe construction for minimal access surgery. Phys Med Biol 2004;49(16):3527–38.