TECHNICAL REPORT

Phantom experiments of axial strain measurements using multidimensional autocorrelation method, multidimensional Doppler method and direct strain measurement method

Chikayoshi Sumi* and Tomonori Ebisawa

*Department of Information and Communication Sciences,
Faculty of Science and Technology, Sophia University,
7-1, Kioicho, Chiyoda-ku, Tokyo, 102-8554 Japan*

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Abstract: Various tissue displacement vector measurement methods have been proposed by us, i.e., the multidimensional cross-spectrum phase gradient method (MCSPGM), the multidimensional autocorrelation method (MAM) and the multidimensional Doppler method (MDM). In this paper, actual measurement results obtained by MAM and MDM are reported. Specifically, for a compressed agar-graphite phantom (infinitesimally compressed, $\approx 0.07\%$; largely compressed, $\approx 0.7\%$), the measurement results for axial strain using MAM and MDM are shown and compared with those using MCSPGM, the conventional one-dimensional (1D) AM and the direct strain measurement method based on 1D AM (DSMM). Although DSMM was also developed by us in 1993, the method and the actual measurement results have not been reported elsewhere. Because the multidimensional methods can provide more accurate axial displacement measurements in real time and enable displacement vector measurement, 1D AM and DSMM might now be ineffective for practical applications.

Keywords: Ultrasound, Axial displacement measurement, Axial strain measurement, Displacement vector, Phantom experiment, Multidimensional autocorrelation method (MAM), Multidimensional Doppler method (MDM), Multidimensional cross-spectrum phase gradient method (MCSPGM), Direct strain measurement method (DSMM)

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1. INTRODUCTION

It is remarkable that the pathological state of human soft tissues highly correlates with static and low-frequency mechanical properties, particularly shear elasticity [1–3]. Accordingly, we have been developing ultrasonic (US)-strain-measurement-based one-dimensional (1D) [1–3], 2D [1–3] and 3D [3,4] shear modulus reconstruction techniques as diagnostic tools for various *in vivo* tissues, such as the breast [5–8] and liver [3,9]. In these respective techniques, a 1D (axial) displacement field, and 2D and 3D displacement vector fields (i.e., generated by compression, vibration, heart motion etc) are measured to obtain 1D (axial) strain, and 2D and 3D strain tensor fields by partial differentiation. Many other researchers are also developing shear modulus reconstruction methods (e.g., [10–13]) based on various displacement/strain measurement methods (e.g., conventional 1D Doppler method and autocorre-

lation method (AM) for blood flow measurement [14] in [15], and crosscorrelation method in [16–18]). Other groups have also used 1D AM to measure axial tissue displacement/strain (e.g., [19]).

As displacement measurement methods, we developed the multidimensional cross-spectrum phase gradient method (MCSPGM) [20,21], the multidimensional autocorrelation method (MAM) [22,23] and the multidimensional Doppler method (MDM) [22,23]. These methods enable real-time displacement vector measurement. To increase the measurement accuracy of the displacement vector and axial displacement component, we developed the multidirectional synthetic aperture method (MDSAM) [3,23,24] and lateral modulation methods such as the lateral Gaussian envelope cosine modulation method (LGECMM) [22–24]. The order of the accuracies (signal-to-noise ratios, SNRs) of the multidimensional methods was investigated for displacement vector measurement by simulations in [23] in the cases of using and not using MDSAM and LGECMM, i.e., for high-echo-data SNR (∞ , 50–30 dB),

*e-mail: c-sumi@sophia.ac.jp

Table 1 Comparisons of MCSPGM, MAM, MDM, 1D AM, and DSMM (based on 1D AM).

Method	Data measured	Phase used	Impossible by aliasing?	Impossible by lateral motion?	Phase matching (PM)
MCSPGM	Displacement vector	Local phase in multidimensional space (MS)	Yes	No	PM should be performed in MS
MAM and MDM	Displacement vector	Instantaneous phase in MS	Yes	No	PM should be performed in MS
1D AM	Axial displacement	Instantaneous phase in axial direction (AD)	Yes	Yes	PM must be performed in lateral direction (LD) and also should be performed in AD
DSMM	Axial strain	Instantaneous phase in AD	No	Yes	PM must be performed in LD and also should be performed in AD

MAM > MCSPGM > MDM. The echo SNRs decreasing, the order changes as follows, MAM > MDM > MCSPGM (20 dB), MDM > MAM > MCSPGM, MDM > MCSPGM > MAM (low-echo-data SNR, 0.7 dB).

Because actual measurement results for MAM and MDM have not been reported elsewhere except in proceedings (e.g., [25]), in this paper, using a compressed agar-graphite phantom (infinitesimally compressed, $\approx 0.07\%$; largely compressed, $\approx 0.7\%$), the measurement results of axial strain using MAM and MDM are shown and compared with those of MCSPGM [20,21], conventional 1D AM [14] and DSMM [26] developed by us (i.e., a direct strain measurement method based on conventional 1D AM). Generally, except for the method using spectrum cross-correlation [27] and DSMM, a strain tensor is obtained from the measured displacement vector by partial differentiation as mentioned above. Although DSMM was described in our Japanese Patent (1993) [26], the method and corresponding measurement results have also not been reported elsewhere. In the appendix, DSMM is described, and in Table 1, MAM, MDM, MCSPGM, 1D AM and DSMM are compared in terms of the type of data measured, the phase used for the measurement, whether aliasing of the phase (i.e., due to large tissue motion) and lateral tissue motion make the measurement impossible, and if so, whether our previously developed phase matching method [3,20,21,23] should be performed. The necessity of phase matching is also reviewed in the appendix.

2. PHANTOM EXPERIMENTS

First, measurement results of axial strain using MCSPGM, MAM, MDM, DSMM and 1D AM are shown and compared using an infinitesimally compressed agar-graphite phantom. An infinitesimal axial strain was mechanically applied to the phantom by compressing the

phantom in the axial direction using the ultrasound transducer itself. The height of the phantom was 60 mm and the applied compression was 0.04 mm. Thus, the mean of the generated axial strain in the phantom is approximately 0.07% (i.e., infinitesimal strain). The phantom had a cylindrical stiff inclusion (dia. = 15 mm) at a depth of 15.0 mm (agar, 7% and 3%, respectively, for inclusion and surrounding region; graphite, 3% for both regions). The relative shear modulus was evaluated by a load examination to be 3.33 (2.96 vs 0.89×10^6 N/m²) for relatively large strains of 0.2 to 0.7%. The ultrasound frequency used was 7.5 MHz and the beam pitch was 0.2 mm (SSD5500, Aloka, Tokyo, Japan). The region of interest (ROI) with an area of 23.8 mm (axial) $\times$ 19.4 mm (lateral) was set at a depth of 3.2 mm. The moving-average widths for MAM [23], MDM [23], DSMM and 1D AM are 0.8 mm (axial) $\times$ 1.8 mm (lateral), and the local region size for MCSPGM is also 0.8 mm $\times$ 1.8 mm.

In Fig. 1, B-mode image of ROI is shown in (a) and the axial strains measured by the methods are shown as linear gray-scale images with the same dynamic range from -0.15 to 0% [(b) MCSPGM, (c) MAM, (d) MDM, (e) 1D AM and (f) DSMM]. When the phantom was mechanically compressed in the axial direction and the generated lateral displacements were less than 0.1 mm (a half beam pitch), the cross-correlation method for coarse phase matching [20,21,23] yielded zero lateral displacement.

As shown, although the axial strain images are stably obtained using MCSPGM, MAM and 1D AM, DSMM yielded a slightly unstable image. The MDM result is also unstable. Recall that for high-echo-data SNR, the order of the accuracies (SNRs) of the multidimensional methods was obtained by simulations [23] and is given in the introduction. For the multidimensional and 1D methods, the means and standard deviations (STDs) evaluated in the inclusion and the surrounding region are summarized in

Fig. 1 Infinitesimally compressed case (mean strain, 0.067%) in phantom experiment. (a) B-mode image. Axial strain images obtained using (b) 2D cross-spectrum phase gradient method (CSPGM), (c) 2D autocorrelation method (AM), (d) 2D Doppler method (DM), (e) 1D AM and (f) direct strain measurement method (DSMM).

Table 2 Means, standard deviations (STDs), ratios of mean strains and CNRs obtained for axial strains measured on an agar-graphite phantom having a stiff inclusion using MCSPGM, MAM, MDM, 1D AM and DSMM in infinitesimally compressed case (mean strain, 0.067%, Fig. 1).

Method	Mean [%]	STD [%]	Ratio	CNR
MCSPGM	-0.0321, -0.0850	0.0303, 0.0444	2.64	1.94
MAM	-0.0489, -0.0781	0.0289, 0.0485	1.60	0.53
MDM	-0.0487, -0.0803	0.1529, 0.1013	1.65	0.06
1D AM	-0.0486, -0.0778	0.0389, 0.0494	1.60	0.43
DSMM	-0.0451, -0.0802	0.0451, 0.0671	1.78	0.38

Table 2. The STDs of MDM and DSMM are larger than those of other methods [e.g., compared with MAM: inclusion, 0.1529 and 0.0451 vs 0.0289%; surrounding, 0.1013 and 0.0671 vs 0.0485%], although the means are almost the same as those of other methods [e.g., inclusion, -0.0487 and -0.0451 vs -0.0489%; surrounding, -0.0803 and -0.0802 vs -0.0781%]. Note that for each method, the STD of the surrounding region, where there is a large strain, is larger than that of the inclusion, where

there is a small strain. However, from all the strain images, the inclusion was quantitatively evaluated to be stiff compared with the surrounding region. The ratios of the mean strains of the inclusion and the surrounding region are also shown in Table 2 (≈ 1.6 to 1.8). Note that for MCSPGM, the ratio of mean strains is large (2.64). The differences between the results are due to reconstruction errors that originate from strain measurement errors. However, all the strain ratios of smaller than 3.33 are caused by the nonlinearity of the target shear modulus and an artifact of 1D reconstruction (strain ratio) [3].

Regarding the detectability of the inclusion in the strain images, using the means and STDs of strains in the inclusion and the surrounding region, the contrast-to-noise ratios (CNRs) [28] are also evaluated for the strain images, as shown in Table 2. The order of the CNRs is MCSPGM > MAM > 1D AM > DSMM > MDM (1.94 to 0.06). For the multidimensional methods, the order is almost the same as that of the SNRs for high-echo-data SNR obtained by simulations [23].

Note that 1D AM yielded STDs and a CNR comparable with those of MAM (inclusion, 0.0389 vs 0.0289%; surrounding, 0.0494 vs 0.0485%; 0.43 vs 0.53). This is

Fig. 2 Largely compressed case (approximately 0.7%) in phantom experiment using the same phantom as that for Fig. 1. Axial strain images obtained using (a) 2D CSPGM, (b) 2D AM, (c) 2D DM, (d) 1D AM, and (e) DSMM.

because the generated lateral displacements are small. However, strictly speaking, even in such a case, MAM yields higher accuracy and detectability than 1D AM because MAM does not disregard the existence of lateral displacements. Moreover, DSMM is inferior to 1D AM (STDs of inclusion, 0.0451 vs 0.0389%; surrounding, 0.0671 vs 0.0494%; CNR, 0.38 vs 0.43).

Next, the corresponding axial strain images obtained by the above methods are shown in Figs. 2(a)–2(e) when sufficiently large strains were generated (approximately 0.7%). Also here, the cross-correlation method yielded zero lateral displacement. Obviously, the detectability of the images obtained by MCSPGM, MAM, MDM and 1D AM [Figs. 2(a)–2(d), respectively] are superior to those in the corresponding infinitesimally compressed case [Figs. 1(b)–1(e)]. The CNRs are summarized in Table 3 together with the means, STDs and ratios of mean strain. The CNRs improved from 0.06–1.94 to 5.11–7.61. This is because the increase in contrast (due to the nonlinearity of the target shear modulus) was larger than that in STD (see in Tables 2 and 3, ratios of mean strains, 1.60–1.65 except for MCSPGM to 2.28–2.55 including MCSPGM and STDs). Note that the order of the CNRs of the multidimensional methods is the same as that of the SNRs for high-echo-data

Table 3 Means, STDs, ratios of mean strains and CNRs obtained for axial strains measured on an agar-graphite phantom having a stiff inclusion using MCSPGM, MAM, MDM, 1D AM and DSMM in largely compressed case (approximately, 0.7%, Fig. 2).

Method	Mean [%]	STD [%]	Ratio	CNR
MCSPGM	−0.293, −0.669	0.0861, 0.2159	2.28	5.23
MAM	−0.268, −0.681	0.0749, 0.1985	2.55	7.61
MDM	−0.283, −0.674	0.1274, 0.2085	2.38	5.11
1D AM	−0.284, −0.682	0.0848, 0.1946	2.40	7.02
DSMM	−0.279, −1.297	0.3480, 4.5780	4.66	9.90×10^{-4}

SNR obtained by simulations [23], i.e., MAM (7.61), MCSPGM (5.23), MDM (5.11). The 1D AM yielded slightly less accuracy and detectability than MAM, similarly to the infinitesimally compressed case in terms of STD (inclusion, 0.0848 vs 0.0749%; surrounding, 0.1946 vs 0.1985%) and CNR (7.02 vs 7.61).

However, for DSMM, the detectability significantly decreased [0.38 from Fig. 1(f) vs 9.90×10^{-4} from Fig. 2(e)]. Although the contrast in strains was enhanced, the decorrelation noise caused by the lateral displacement was also enhanced by the large compression, i.e., the

generated noise was amplified by a factor of 2 by the differentiation (STD of inclusion, 0.3480%; that of surrounding area, 4.5780%). Thus, for DSMM, strictly speaking, phase matching should be performed using the analogue estimate of axial and lateral displacements (i.e., by MCSPGM using thinned-out echo data [3,23]). Otherwise, lateral echo data should be interpolated through a discrete Fourier transform.

For small strains in the infinitesimally compressed case and those in the stiff inclusion in the largely compressed case, the order of the accuracies (STDs) of the multidimensional methods is almost the same as that of the SNRs for high-echo-data SNR obtained from simulations [23]. The large strain is also a source of decorrelation noise. Thus, for the large strains in the surrounding region in the largely compressed case, the order of STDs is the same as that of the SNRs for slightly low echo data SNR (20 dB in simulations [23]), i.e., $\text{MAM} > \text{MDM} > \text{MCSPGM}$. The decorrelation noise also affected the result obtained by DSMM.

In Figs. 1 and 2, the stiff inclusion on the phantom cannot be visualized clearly in the lateral strain images due to the inaccuracy of the measurements (omitted). Visualized cases are shown, for instance, in [20,29]. As shown, because neither MDSAM [3,23] nor lateral modulation methods [22–24] are used, the lateral displacement measurement is unstable. Although we have also reported the effectiveness of a regularization on stabilization of the lateral displacement measurement [29–31] (regularization more effectively suppresses the measurement noise than low-pass filtering and can also be applied to axial displacement), the effectiveness of these methods is outside the scope of this paper.

3. CONCLUSIONS

We carried out actual axial strain measurements using MAM and MDM. The measurement accuracies (STDs) and detectabilities (CNRs) of the axial strain measurements performed by the methods were also compared with those of our previously developed MCSPGM, a direct strain measurement method based on 1D AM (DSMM) and conventional 1D AM. The measurement accuracy of DSMM was lower than those of the multidimensional methods and 1D AM, particularly for large strains.

As confirmed, for the measurement accuracy of the large strains ($\approx 0.7\%$), the order of the accuracies of the multidimensional methods was the same as that of the SNRs for slightly low echo data SNR (20 dB in simulations [23]). However, for small strains of less than 0.3%, the order of the accuracies of the multidimensional methods was almost the same as that of the SNRs for high-echo-data SNR (∞ , 50–30 dB in simulations [23]). For such strains, the order of the detectabilities of the multidimensional

methods was also almost the same as that of the SNRs for high-echo-data SNR. Because the multidimensional methods can provide more accurate axial displacement measurements than 1D AM and DSMM and enable displacement vector measurement in real time, 1D AM and DSMM might now be ineffective for practical applications.

In this study, the lateral strain measurements were inaccurate. The effectiveness of MDSAM [3,23] and lateral modulation methods [22,23] on improving measurement accuracy will be reported elsewhere together with that of a regularization method [29–31]. These methods will enable accurate multidimensional displacement vector measurement and multidimensional shear modulus reconstruction [3,4,29,32].

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Appendix: DSMM based on 1D AM

Here, we describe DSMM [26]. On the basis of 1D AM [14,15], displacement in the axial (US beam) direction x is expressed as

$$u_x(x) = -\frac{c}{4\pi f} \theta(x), \quad (1)$$

where $\theta(x)$ is the phase of the autocorrelation signal $AC(x)$ (i.e., a spatial correlation signal with no lag obtained from the analytic signal or quadrature signal sampled with a pulse repetition interval Δt), c is the ultrasound speed and f is the ultrasound frequency. Axial strain is obtained by differentiating the axial displacement $u_x(x)$ in the axial direction. Thus, the axial strain $\varepsilon_{xx}(x)$ is expressed as

$$\begin{aligned} \varepsilon_{xx}(x) &= \frac{\partial}{\partial x} u_x(x) \\ &= -\frac{c}{4\pi f} \frac{\partial}{\partial x} \theta(x) \\ &= -\frac{c}{4\pi f} \frac{\operatorname{Re}[AC(x)] \times \frac{d}{dx} \operatorname{Im}[AC(x)] - \frac{d}{dx} \operatorname{Re}[AC(x)] \times \operatorname{Im}[AC(x)]}{\operatorname{Re}[AC(x)]^2 + \operatorname{Im}[AC(x)]^2}. \end{aligned} \quad (2)$$

Using Eq. (2), $\varepsilon_{xx}(x)$ is directly calculated by the four rules of arithmetic using the real and imaginary components of $AC(x)$ ($\text{Re}[AC(x)]$ and $\text{Im}[AC(x)]$) and their derivatives (obtained using a differential filter with a cutoff frequency [2]) without calculating the phase $\theta(x)$. This enables us to directly obtain axial strain data without unwrapping the phase when the displacement is larger than $\lambda/4$. Also as shown for MAM [22,23], to obtain $\theta(x)$ in 1D AM [Eq. (1)], the spatiotemporal moving average should be applied to $\text{Re}[AC(x)]$ and $\text{Im}[AC(x)]$. This method of signal processing yields more accurate measurement results than calculating the moving average of the raw phase. Correspondingly, in DSMM, the moving average is also applied to $\text{Re}[AC(x)]$ and $\text{Im}[AC(x)]$ in Eq. (2). However, in the phantom experiments, the spatial moving average is used instead of the spatiotemporal moving average so that the temporal resolution does not decrease.

DSMM can be combined with our previously reported 2D [20,21] or 3D [3] phase-matching method to remove the lateral displacement that lowers the accuracy of axial displacement measurement owing to the occurrence of echo decorrelation [33], where the cross-correlation method [16,17] is used to roughly estimate the large displacement vector by detecting the peak position of the digital crosscorrelation function. That is, after shifting local echo data using the obtained rough estimate, the remaining displacement is estimated, i.e., the rough estimate is updated by adding the estimate of the remaining displacement. Originally, the phase-matching method was used to allow for decorrelation and aliasing caused by the generation of a large displacement when MCSPGM was used [20,21]. This phase-matching method can also be used in advance when MAM and MDM are used [22,23]. By

thinning out local echo data [3], MCSPGM can also be used instead of the cross-correlation method as a rough estimator. Also in the application of conventional 1D AM in [19], the phase-matching method was used, which enables the manual measurement [34] of axial displacement and axial strain for *in vivo* soft tissues.

When the autocorrelation signal is obtained not from quadrature signals but from analytic signals, as in MAM [22,23], the coefficient $-c/(4\pi f)$ in Eqs. (1) and (2) can be estimated by differentiating the phase of the spatial correlation signal along the axial direction. This yields a robust displacement measurement with respect to the frequency modulation [35] caused by attenuation and scattering. This method of signal processing was used in this study.

The feasibility of DSMM is also verified by a simple simulation in [36]. 1D echo data were simulated by convolving the 1D Gaussian envelope point spread function (ultrasound speed, 1,500 m/s; ultrasound frequency, 3.5 MHz; λ , 4×0.107 mm) with white noise. Echo noise was not added. The applied strain was 0.5 percent (elongation) with a body motion of 4.31×10^{-2} mm. Although aliasing occurred using 1D AM at the position where the displacement reaches approximately 0.107 mm (Fig. 1(a) in [36]), axial strain was directly measured using DSMM without unwrapping the phase (Fig. 1(b) in [36]). However, the DSMM result is slightly less stable than the 1D AM result (STDs, 0.064 vs 0.048%; means, 0.503 vs 0.489%), although mathematically, the same axial strain is calculated by Eq. (2). This reveals that DSMM is very sensitive to the noise existing in the analytic signals (i.e., even computational errors, etc). Such high sensitivity is also shown for the experimental echo data, i.e., for cases obviously including measurement noises.