Performance Evaluation of Coherence-Based Adaptive Imaging Using Clinical Breast Data

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Abstract—Sound-velocity inhomogeneities degrade both the spatial resolution and the contrast in diagnostic ultrasound. We previously proposed an adaptive imaging approach based on the coherence of the data received in the channels of a transducer array, and we tested it on phantom data. In this study, the approach was tested on clinical breast data and compared with a correlation-based method that has been widely reported in the literature. The main limitations of the correlation-based method in ultrasonic breast imaging are the use of a near-field, phase-screen model and the integration errors due to the lack of a two-dimensional (2-D) array. In contrast, the proposed coherence-based method adaptively weights each image pixel based on the coherence of the receive-channel data. It does not make any assumption about the source of the focusing errors and has been shown to be effective using 1-D arrays. This study tested its *in vivo* performance using clinical breast data acquired by a programmable system with a 5 MHz, 128-channel linear array. Twenty-five cases (6 fibroadenomas, 10 carcinomas, 6 cysts, and 3 abscesses) were investigated. Relative to nonweighted imaging, the average improvements in the contrast ratio and contrast-to-noise ratio for the coherence-based method were 8.57 dB and 23.2%, respectively. The corresponding improvements when using the correlation-based method were only 0.42 dB and 3.35%. In an investigated milk-of-calcium case, the improvement in the contrast was 4.47 dB and the axial and lateral dimensions of the object were reduced from 0.39 to 0.32 mm and from 0.51 to 0.43 mm, respectively. These results demonstrate the efficacy of the coherence-based method for clinical ultrasonic breast imaging using 1-D arrays.

I. INTRODUCTION

A PRIMARY performance limitation of diagnostic ultrasound is the distortion introduced by sound-velocity inhomogeneities in human tissue. Because the sound velocity varies over a wide range (e.g., 1450 m/s in fat and 1665 m/s in collagen [1]), the use of a constant sound velocity when calculating focusing delays inevitably results in focusing errors. Such focusing errors also are termed phase aberrations, and they degrade both the spatial resolution and the contrast in the resultant images.

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Several phase-aberration correction techniques have been proposed to adaptively reduce these errors and thereby improve the image quality [2]–[15]. Flax and O'Donnell [2] proposed a correlation-based method that modeled sound-velocity inhomogeneities as a near-field phase screen, which assumes that time-delay errors produce phase aberrations only at the surface of the transducer. The near-field phase screen can be estimated by finding the peak of the cross-correlation function of the signals from adjacent channels [2], [3], or using those from the individual channels to the beam sum [4] or partial beam sum [5]. Phase aberrations theoretically can be corrected by compensating for the time-delay errors, but several factors have limited the clinical applications of correlation-based methods. First, the near-field phase-screen model is often inadequate because sound-velocity inhomogeneities result in not only time-delay errors but also energy-level fluctuations and waveform distortion. This led to a displaced phase-screen model that accounts for phase aberrations as a phase screen positioned some distance from the transducer surface [6]. However, its performance in ultrasonic breast imaging is still insufficient because the sound-velocity variations in breast are more distributed. The other reason for the inadequate performance of correlation-based methods is the three-dimensional (3-D) nature of sound-velocity inhomogeneities. Preventing aberrator integration errors in time-delay estimation requires the use of a 1.5-D or 2-D array [7], [16]. However, these are not yet in widespread use in clinics due to the significantly increased system complexity and cost.

Adaptive sidelobe-reduction techniques also can be used to reduce focusing imperfections [8]–[15], one example of which is the parallel adaptive receive-compensation algorithm (PARCA) [8]–[11]. At each image point, the PARCA removes unwanted sidelobe contributions and maintains the mainlobe signal intensity by estimating the on-axis and off-axis signal components. However, a significant drawback of PARCA is its computational complexity.

Another adaptive sidelobe-reduction method is based on the use of the coherence factor (CF) of the receive-channel data [12]–[14], which represents a focusing-quality index:

$$CF = \frac{\left| \sum_{i=0}^{N-1} S(i, t) \right|^2}{N \sum_{i=0}^{N-1} |S(i, t)|^2}. \quad (1)$$

where N denotes the total number of array channels, $S(i)$ is the signal received at channel i , and t denotes the time corresponding to the depth of the image point. In other words, the CF is defined as the ratio between the energy of the coherent sum to the total incoherent energy (i.e., N times the incoherent sum). Therefore, an image point with a high CF value possesses good focusing quality, and its amplitude should be maintained, whereas the amplitude of an image point with a low CF value possesses poor focusing quality and its amplitude should be suppressed. Thus, this method is effectively an adaptive weighting technique in which the amplitude of each image pixel is weighted by the corresponding CF such that the unwanted sidelobes are effectively reduced. The CF also has been modified to the generalized CF (GCF) to accommodate the coherence properties of speckle-generating objects [15]. The efficacy of the GCF-weighting method has been demonstrated for phantom data [15]. The main purpose of the present study was to test the feasibility of applying the GCF to clinical breast data. Specifically, the method was applied to different types of breast lesions: cyst, fibroadenoma, carcinoma, and abscess.

This paper explores the contrast and other general characteristics of the GCF technique. It is organized as follows. In Section II, the basic principles and experimental methods for the GCF-weighting method are described. Section III presents the experimental results. The results then are discussed in Section IV. The paper concludes in Section V.

II. IMAGING PRINCIPLES AND RESEARCH METHODS

A. Generalized Coherence Factor

The GCF is calculated using the Fourier spectrum of the aperture-domain data, which are the data received in individual array channels after the application of focusing delays but prior to beam summation. The corresponding Fourier spectrum is obtained by performing a discrete Fourier transform across the array at each time (i.e., depth):

$$S(i, t) \xrightarrow{\text{FT}} p(k, t), \quad (2)$$

where $p(k, t)$ is the Fourier spectrum with spatial frequency index k and time variable t . As described in [11], the spectrum can be viewed as an approximation of multiple parallel receive beams centered at the transmit beam direction. The definition of the CF described in (1) also can be expressed as follows:

$$\text{CF} = \frac{|p(0, t)|^2}{\sum_{k=-N/2}^{N/2-1} |p(k, t)|^2}. \quad (3)$$

The numerator of (3) is the energy of the dc component of the spectrum, and the denominator is the total

spectral energy. The dc component represents the signals received from the transmit beam direction (i.e., on-axis component). The other components represent the signals from other angles (i.e., off-axis components).

Adaptive weighting using CF is feasible for point-like targets. However, for diffuse scatterers (i.e., speckle-generating objects), part of the mainlobe signal appears within the low-frequency region. Therefore, using the ratio of the dc energy to the total energy as a weighting factor underestimates the coherence. The GCF was proposed to take this into account [15], and it thus is defined as the ratio of the energy within a certain low-frequency region to the total energy:

$$\text{GCF} = \frac{\sum_{k=-M_0}^{M_0} |p(k, t)|^2}{\sum_{k=-N/2}^{N/2-1} |p(k, t)|^2}, \quad (4)$$

where M_0 denotes the prespecified cut-off frequency in the spatial-frequency domain. Note that the CF is the special case of the GCF with $M_0 = 0$.

B. Signal Processing Procedures

The signal processing procedures we used are shown schematically in Fig. 1. After the echo signal is received and digitized by the A/D converter, the received RF data are demodulated down to baseband, and the baseband beamformer applies geometric focusing delays and phase rotations to the baseband data. The B-mode image then is obtained by performing beam summation across the array with prespecified apodization. Note that the geometric delays are adjusted for every depth on receive to achieve dynamic focusing. However, as previously mentioned, the focusing delays applied in a beamformer are derived by assuming a constant sound velocity without considering the sound velocity variations in soft tissues. Therefore, it is the purpose of the proposed GCF method to estimate and suppress such focusing errors. Because the carrier frequency needs to be removed when calculating the coherence factor, baseband beamformers are preferred. The GCF is calculated for the specific configuration using the delayed baseband data across the aperture at all ranges prior to beam summation. The amplitude of the beam sum then simply is multiplied by the corresponding GCF:

$$x_{\text{weighted}} = \text{GCF} \cdot x, \quad (5)$$

where x denotes the original amplitude of the beam-sum data, and x_{weighted} denotes the amplitude of the beam sum weighted by the GCF. The weighted beam-sum data then are stored in a buffer for further signal processing, scan conversion, and display.

C. Clinical Data Acquisition

Experiments were performed to evaluate the efficacy of the proposed method in clinical breast imaging. Clinical

Fig. 1. Signal processing procedures of the proposed GCF-weighting technique.

Fig. 2. Clinical data-acquisition apparatus.

data of the female breast were collected in the ultrasonic consulting room of the Taipei Veterans General Hospital by one of the authors (Y.-H. Chou) after patients had provided informed consent. A total of 25 patients aged from 23 to 89 years were included. Among them, there were 6 fibroadenomas, 10 carcinomas, 6 cysts, and 3 abscesses. A schematic of the experimental apparatus is shown in Fig. 2, which included a mechanical compression stage similar to that used in X-ray mammography to stabilize the breast during data acquisition. A programmable imaging system with 64 system channels (DiPhAS, IBMT, Fraunhofer Institutes, St. Ingbert, Germany) was used to acquire the aperture-domain data (i.e., the receive-channel data prior to beam summation). The linear transducer array had a center frequency of 5 MHz, a -6 dB bandwidth of 4.1 MHz and 128 channels (L6/128, Sound Technology, State College, PA) at a pitch of 0.3 mm. A complete data set was obtained from each patient (i.e., 128×128 RF A-mode scans recorded from the 128-element array, for which each A-mode scan represents the echo signal of a particular transmit/receive combination) at a sampling rate of 20 MHz. Each complete data set took approximately 3 s to acquire. The acquired data sets were used to reconstruct the aperture-domain data with full dynamic focus on both transmit and receive. In order to evaluate tissue motion, B-mode images were acquired and stored before and after

full data set acquisition. The two images then were compared to make sure that no noticeable tissue motion was present. An f-number of 1 was applied for aperture control. All images displayed here are in linear gray scale mapping with a dynamic range of 60 dB and are shown in a linear scan format. The step size for the linear scan was 0.15 mm (i.e., half the pitch of the transducers). B-mode images are formed by applying transmit and receive focusing delays before beam summation, and multiple imaging methods can be compared with one another using the same data set.

III. EXPERIMENTAL RESULTS

A. Cut-off Frequency M_0

In the GCF technique, cut-off frequency M_0 should be specified so as to achieve optimal performance. Both the contrast ratio (CR) and contrast-to-noise ratio (CNR) were used in the experiments to evaluate the effects of varying M_0 . The CR is defined as the ratio between the mean value in a background region to that in a region of interest (ROI) [3]. The CNR is defined as the ratio between the CR and the standard deviation of the image intensity in a background region [4]. Fig. 3(a) shows an original image without application of the GCF-weighting method, and Figs. 3(b)–(f) show the GCF-weighted images with M_0 ranging from 0 to 4. Note that $M_0 = 0$ represents the case of using the CF. The background region and ROI used to calculate CRs and CNRs are indicated in Fig. 3(a) by the black and white boxes, respectively. The ROI in this case is a cyst. Note that in all cases the contrast is clearly improved, and the background noise is suppressed. Figs. 4(a) and (b) show the CR and CNR after GCF weighting as functions of M_0 , respectively. Fig. 4(a) shows that the CR was improved in all cases, decreasing as M_0 increased. When the background-intensity variations are taken into account, Fig. 4(b) shows that the CNR was not improved when $M_0 = 0$. Nevertheless, with $M_0 \geq 1$, CNR improvement is again demonstrated. Specifically, the optimal CNR was achieved at $M_0 = 3$ in this case. The results indicate that using CF as a weighting index can

Fig. 3. Effects of cut-off frequency M_0 on the images of a cyst, displayed with a dynamic range of 60 dB. (a) Original image. Black box indicates background region. White box indicates ROI. (b)–(f) GCF-corrected images for M_0 values of 0 (b), 1 (c), 2 (d), 3 (e), and 4 (f).

effectively increase the CR. However, CF also gives rise to strong intensity variations and degrades the CNR. This also is illustrated in Fig. 3(b) in which the CF-weighting technique introduced artifactual black holes in the background region, which in turn affect the detection of an ROI. Based on these results, $M_0 = 3$ was used in the subsequent investigations (which are described below).

B. Cyst

The sonogram needs to be interpreted accurately in order to avoid unnecessary biopsies and to differentiate a cyst from a solid mass. Fig. 5 shows images of a cyst in a

66-year-old female patient. The background and the ROI regions used in the CR and CNR calculations are indicated in Fig. 5(a) by the black and white boxes, respectively. Fig. 5(a) shows the original image, Fig. 5(b) shows the image obtained using the correlation-based method (as a reference) [2], [3], and Fig. 5(c) is the image obtained with adaptive compensation using GCF weighting. For the correlation-based method, time delay between adjacent channels was calculated by finding the peak position of the cross-correlation function of the signals received by the two channels. An infinite isoplanatic patch size (i.e., single aberration profile applied throughout the image) was used in this study without smoothing the aber-

Fig. 4. Values of CR (a) and CNR (b) for the specimen shown in Fig. 3 as functions of M_0 .

Fig. 5. Images of a cyst in a 66-year-old woman, displayed with a dynamic range of 60 dB: original B-mode image (a), and images obtained using the correlation-based method (b) and the GCF-weighting method (c). The black box in (a) indicates the background image. The white box indicates the ROI. (d) Intensities of the scan-line data along the gray dotted lines shown in (a) and (c) using solid and dashed lines, respectively.

ration profile. The aberration profile is applied after the linear components were removed. The CRs and CNRs were 20.97 dB and 3.07 for the original image, 20.52 dB and 2.95 for the correlation-based method, and 33.53 dB and 4.33 for the GCF-weighting method. With the correlation-based method, the CR and CNR were slightly degraded, by 0.44 dB and 3.85%, respectively. This may be due to the phase aberrations in the breast being more distributed, and aberrator integration errors may be present with the 1-D array. In contrast, the improvements in the CR and CNR using the GCF-weighting method were 12.56 dB and 40.97%, respectively. The original image shows an indistinct internal echo within the cystic mass. The GCF-weighted image shows that the signal within the cystic mass was effectively suppressed and that the CR was enhanced. The solid line in Fig. 5(d) shows the intensity along the scan line shown as the dotted vertical line in Fig. 5(a). With the proposed method, the intensity along this line [i.e., the vertical dotted line in Fig. 5(c)] is shown as the dashed line in Fig. 5(d). Note that image intensities between 1–1.4 cm and 2.2–2.5 cm are similar in the original image [Fig. 5(a)]. With the application of the proposed method; however, image intensities resulting from unwanted sidelobes in the cyst region (2.2–2.5 cm) are effectively reduced. The image intensities in the tissue region (1–1.4 cm) are less affected. Therefore, effective sidelobe reduction is demonstrated, and it is clear that the method is not just a simple “reject” control. The margin of the cyst was more distinct than that in the original image. These results illustrate the significant improvement in contrast and boundary definition in this case.

C. Fibroadenoma

Fig. 6 shows images of a fibroadenoma lesion in a 42-year-old female patient. The background and the ROI regions used in the calculation of the CR and CNR are indicated in Fig. 6(a) by the black and white boxes, respectively. Fig. 6(a) shows the original image, and Figs. 6(b) and (c) show the corrected images obtained using the correlation-based method and the GCF-weighting method, respectively. The CRs and CNRs were 16.92 dB and 2.76 for the original image, 17.51 dB and 2.82 for the correlation-based method, and 25.73 dB and 3.67 for the GCF-weighting method. The CR and CNR improved by 0.58 dB and 2.36% with the correlation-based method and by 8.81 dB and 33.07% with the GCF-weighting method. The original image shows poor boundary definition. However, with the application of the correlation-based method, no noticeable change was observed, and the method was not able to improve the image. But, the use of the GCF-weighting method noticeably improves the contrast and boundary definition in this case. In spite of these improvements, the suppression of the echo signal in the ROI region reduces the delineation of the interior echo, resulting in possible confusion when differentiating a fibroadenoma (solid) from a cyst (liquid). This requires further investigation. Nonetheless, in this case the lesion is determined as

a fibroadenoma without ambiguity based on the boundary echogenicity. Typically, a cyst has brighter echoes at the boundary, and those of a fibroadenoma are not as strong.

D. Carcinoma

Fig. 7 shows images of a carcinoma lesion in a 63-year-old female patient. The background and the ROI used in the calculation of the CR and CNR are indicated in Fig. 7(a) by the black and white boxes, respectively. Fig. 7(a) shows the original image, and Figs. 7(b) and (c) show the corrected images obtained using the correlation-based method and the GCF-weighting method, respectively. The CRs and CNRs were 13.63 dB and 2.19 for the original image, 13.80 dB and 2.23 for the correlation-based method, and 21.52 dB and 2.87 for the GCF-weighting method. The CR and CNR improved by 0.16 dB and 1.59% with the correlation-based method and by 7.89 dB and 30.75% with the GCF-weighting method, respectively. Again, the original image shows an indistinct lesion margin. The internal echo and several echogenic foci indicating microcalcifications within the breast parenchyma also are not clearly evident. However, with the application of the correlation-based method, no noticeable change was observed, and the method was not able to improve the image. The GCF-weighted image shows better characterization of the margin, enhanced delineation of the internal architecture, and enhanced conspicuity of microcalcifications within the breast parenchyma due to improved spatial resolution and contrast.

E. Abscess

Fig. 8 shows images of an abscess lesion in a 31-year-old female patient. The background and the ROI used in the calculation of the CR and CNR are indicated in Fig. 8(a) by the black and white boxes, respectively. Fig. 8(a) shows the original image, and Figs. 8(b) and (c) show the corrected images obtained using the correlation-based method and the GCF-weighting method, respectively. The CRs and CNRs were 22.20 dB and 3.73 for the original image, 22.85 dB and 3.80 for the correlation-based method, and 32.11 dB and 4.59 for the GCF-weighting method. The CR and CNR improved by 0.65 dB and 1.83% with the correlation-based method and by 9.91 dB and 23.10% with the GCF-weighting method, respectively. The original image shows a well-circumscribed hypoechoic mass with internal echogenic debris, a slight posterior acoustic enhancement, and indistinct calcification. However, with the application of the correlation-based method, no noticeable change was observed, and the method was not able to improve the image. The GCF-weighted image shows a clearer lesion margin and enhanced delineation of the internal architecture due to improvement of the contrast.

F. Milk of Calcium

Fig. 9 shows images of a milk of calcium in a cyst lesion in a 25-year-old female patient. Fig. 9(a) shows the original image, and Figs. 9(b) and (c) show the corrected

Fig. 6. Images of a fibroadenoma in a 42-year-old woman, displayed with a dynamic range of 60 dB: original B-mode image (a), and images obtained using the correlation-based method (b) and the GCF-weighting method (c). The black box in (a) indicates the background region. The white box indicates the ROI.

Fig. 7. Images of a carcinoma in a 63-year-old woman, displayed with a dynamic range of 60 dB: original B-mode image (a), and images obtained using the correlation-based method (b) and the GCF-weighting method (c). The black box in (a) indicates the background region. The white box indicates the ROI.

Fig. 8. Images of an abscess in a 31-year-old woman, displayed with a dynamic range of 60 dB: original B-mode image (a), and images obtained using the correlation-based method (b) and the GCF-weighting method (c). The black box in (a) indicates the background region. The white box indicates the ROI.

Fig. 9. Images of a milk of calcium in a cyst in a 25-year-old woman, displayed with a dynamic range of 60 dB: original B-mode image (a), and images obtained using the correlation-based method (b) and the GCF-weighting method (c).

TABLE I
SUMMARY OF THE CR AND CNR VALUES.

Patient no.	Age (years)	Lesion type ¹	CR			CNR			CR improvement (dB)		CNR improvement (%)		Subjective score
			Original	Corr. ¹ based	GCF	Original	CF	GCF	Corr. based	GCF	Corr. based	GCF	
1	53	Cyst	38.84	38.51	53.19	5.84	5.82	6.96	−0.33	14.34	−0.28	19.18	2+
2	49	Cyst	42.00	41.58	57.05	6.09	6.37	7.08	−0.41	15.04	4.51	16.19	2+
3	59	CA	26.00	26.53	36.70	4.03	3.99	4.73	0.52	10.69	−0.88	17.43	1+
4	28	FA	16.88	17.16	23.48	3.05	2.97	3.77	0.28	6.59	−2.73	23.37	1+
5	63	CA	13.63	13.80	21.11	2.19	2.23	2.85	0.16	7.47	1.59	28.09	2+
6	79	CA	10.39	11.05	19.57	1.53	1.59	2.46	0.65	9.17	3.85	60.94	2+
7	66	Cyst	20.97	20.52	33.53	3.07	2.95	4.33	−0.44	12.56	−3.85	40.97	2+
8	43	CA	4.35	4.84	7.67	0.66	0.73	0.96	0.49	3.32	11.65	45.41	1+
9	33	Abscess	14.11	14.57	20.05	2.16	2.23	2.59	0.46	5.93	2.96	19.58	1+
10	59	CA	16.37	17.94	24.87	2.58	2.79	3.29	1.56	8.50	7.92	27.24	2+
11	63	FA	15.32	15.28	20.59	2.67	2.65	2.99	−0.03	5.27	−0.90	12.05	1+
12	46	FA	15.74	16.12	20.94	2.47	2.53	2.89	0.38	5.20	2.30	16.94	1+
13	63	Abscess	12.99	13.05	20.68	2.20	2.15	3.05	0.06	7.68	2.60	38.29	2+
14	89	CA	12.22	12.26	19.43	1.70	1.73	2.04	0.03	7.20	1.90	20.04	1+
15	23	FA	14.19	14.07	18.40	2.54	2.65	2.92	−0.11	4.20	4.28	14.94	1+
16	50	CA	21.07	22.07	30.27	3.55	3.69	4.21	0.99	9.19	3.86	18.36	1+
17	42	FA	16.36	16.91	24.12	2.67	2.73	3.44	0.54	7.75	2.23	28.97	2+
18	25	Cyst	15.45	19.61	27.09	2.56	3.12	3.18	4.16	11.63	22.09	24.50	2+
19	45	CA	21.56	21.18	29.26	3.55	3.29	3.92	−0.37	7.70	−7.26	10.29	1+
20	45	CA	15.85	16.06	21.33	2.98	3.07	3.46	0.21	5.48	3.17	16.17	1+
21	47	Cyst	23.10	23.83	34.91	3.53	3.74	4.08	0.73	11.81	6.02	15.62	1+
22	55	FA	24.18	24.33	33.58	3.80	3.86	4.31	0.14	9.39	1.52	13.30	1+
23	58	Cyst	24.43	24.20	35.70	3.45	3.83	4.32	−0.22	11.26	11.07	25.18	1+
24	31	Abscess	22.20	22.85	30.98	3.73	3.80	4.37	0.65	8.78	1.83	17.12	1+
25	57	CA	26.51	26.92	34.83	4.14	4.33	4.56	0.40	8.31	4.51	10.02	2+

¹CA = carcinoma, FA = fibroadenoma, Corr. = Correlation.

images obtained using the correlation-based method and the GCF-weighting method, respectively. The contrast between the milk of calcium and the tissue improved by 0.05 dB for the correlation-based method and by 4.47 dB for the GCF-weighting method. The lateral and axial -6-dB widths of the milk of calcium object were 0.51 and 0.39 mm in the original image, 0.55 and 0.41 mm in the correlation-based image, and 0.43 and 0.32 mm in the GCF-weighted image, respectively. The GCF-weighting method improved both the spatial resolution and the contrast in this case, which in turn helped in the detection of calcification.

IV. DISCUSSION

The CR and CNR values obtained with the correlation-based and GCF-weighting methods are listed in Table I. The mean values for the CR and CNR improvements were 0.42 dB and 3.4% for the correlation-based method and 8.57 dB and 23.20% for the GCF-weighting method. It should be noted that, although the CR and CNR are widely reported as performance indices in the literature and are directly related to fundamental characteristics of an image, their exact values may change with the selected ROI and may not be directly related to diagnostic out-

comes. Therefore, these images also were evaluated by an experienced clinician (Y.-H. Chou) using the following scoring system: 2+, significant improvement; 1+, moderate improvement; 0, no noticeable effects; 1−, slight degradation; and 2−, significant degradation. In general, application of the GCF-weighting method improved the image quality in all the cases assessed in this study. The improvement in the contrast was more significant for a liquid lesion (e.g., a cyst) than for a solid lesion (e.g., a fibroadenoma), and the improvement in the boundary definition makes it easier to detect a malignant tumor (e.g., a carcinoma).

The proposed method requires only the use of individual channel data for each firing. Nonetheless, complete data sets were acquired in this study. With the complete data set, dynamic, two-way focusing can be realized. Therefore, the problem associated with reduced coherence at positions away from the fixed transmit focus can be avoided. However, acquisition of a complete data set suffers from lower signal-to-noise ratios and potential motion artifacts. In clinical situations, a reasonable compromise is to use multizone focusing on transmit.

Although the correlation-based method is one of the most popular adaptive imaging methods reported in the literature, several problems were encountered when we applied this to breast imaging in this study. The estimated time delays were affected by several factors, including the cumulative error along the array direction, the signal-to-noise ratio, the correlation between adjacent channels, the kernel size, and the aberrator integration errors of a 1-D array [17]–[19]. In addition to the above problems, the distributed sound-velocity inhomogeneities also represent a uniquely difficult problem in adaptive breast imaging, as illustrated in Fig. 10. The image shown in Fig. 10(a) is displayed without any aberration correction. Fig. 10(b) plots the estimated time-delay errors as a function of the channel index at the image points corresponding to the boxes marked in Fig. 10(a), which are located along the same beam and are spaced by 7.7 mm. The time delays were estimated according to previously described principles [3]. Significant variations in the estimated time delays are clearly evident, and these limit the performance of the correlation-based method. The depth dependency of the estimated time delay may result from breast tissue comprising a heterogeneous mixture of fatty, fibrotic, connective, and glandular tissues.

The arrival time and energy level reportedly vary with breast sample thickness [20], and the correlation length of the aberration profile in breast tissue varies with age and body type [20]–[22]. Therefore, it is to be expected that the near-field, phase-screen model is inaccurate, resulting in poor breast imaging when using the correlation-based method. It also has been reported that waveform distortion is severe and sidelobes are high for breast tissue, primarily because it is a highly refracting and heterogeneous medium [20], [23]–[25]. Therefore, adaptive sidelobe-reduction techniques, such as the GCF-weighting method investigated in the study, are of particular interest because they do not make any assumptions regarding the origin of the side-

Fig. 10. Estimated time-delay errors. (a) Clinical image displayed with a dynamic range of 60 dB. (b) Corresponding estimated time-delay errors as functions of channel index at the imaging points indicated in (a).

lobes. As shown in this paper, application of the GCF-weighting method significantly improves the contrast and boundary definition for cyst and solid masses. We also have demonstrated that the GCF-weighting method is more effective than correlation-based methods in improving the quality of images obtained from ultrasonic breast imaging without the use of a 2-D (or 1.5-D) array.

This paper illustrates efficacy in sidelobe reduction of the proposed method using 1-D arrays. Nonetheless, the proposed method also can be extended to imaging with 1.5-D/2-D arrays. In particular, because the aperture also is sampled along the elevational direction, the proposed method can be applied to improve elevational focusing quality. Signal processing required for elevational GCF weighting with 1.5-D/2-D arrays will be similar to that for lateral GCF weighting with 1-D arrays.

V. CONCLUSIONS

In this paper, we have experimentally evaluated an adaptive imaging approach for reducing focusing errors

in ultrasonic imaging. This method is based on coherence and corrects focusing errors by suppressing unwanted side-lobes. The experimental results demonstrate that the image quality was noticeably improved in all of the cases investigated using a 1-D array.

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