Compensation method for frequency-dependent attenuation in tissue imaging by amplitude modulation for chirp transmission

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Abstract: Ultrasound imaging of deep parts in a living body with high resolution and high signal-to-noise ratio is strongly required. The pulse compression technique is efficient for achieving a high signal-to-noise ratio, and wide-band imaging is important for realizing high image resolution. Hence, a wide-band transducer is intensively studied. However, high-frequency components in a wide-band transmission tend to be affected by frequency-dependent attenuation. This decreases not only the signal-to-noise ratio but also the image resolution, since the distortion of echo signals makes exact pulse compression impossible, i.e., the pulse width of the compressed echo signal becomes broad. In this study, we examine the compensation of frequency-dependent attenuation in pulse compression imaging using FM chirp signals.

Keywords: Frequency-dependent attenuation, FM chirp signal, Pulse compression, Attenuation compensation

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1. INTRODUCTION

The goal of our study is to perform high-resolution, and high-signal-to-noise-ratio (SNR) ultrasound imaging to realize high-quality diagnosis. This is strongly required, especially for deep parts in a living body. To improve the resolution and the SNR, the pulse compression technique (PCT) is an effective method that is safe for the living body [1–3].

In the PCT, to improve range resolution, a wide-band modulation is required. The bandwidth that can be used efficiently for transmission and reception is limited by the resonance characteristics of the transducer that utilizes thickness vibration resonance. To broaden the bandwidth, layered-type transducers are being developed, in which two piezoelectric oscillators of different thicknesses are adhered longitudinally and one of a pair of electrodes is inserted between the two oscillators [4]. In general, as the bandwidth of the transmitted pulse is wide, the pulse width of the signal processed by the PCT becomes narrow, and hence, the range resolution is improved [1]. To improve the SNR, a long-period signal should be transmitted to increase the power introduced into the body. Namely, the PCT with the wide-band modulation and the long-period transmission is suitable for our purpose.

Applying the PCT is expected to improve the SNR while retaining the high resolution, but the frequency-dependent attenuation (FDA) must be considered. FDA causes severe distortion of echo signals, when the wide-band pulse propagates through soft tissue in a living body. The distortion of echoes caused by FDA makes exact pulse compression impossible, and hence, image blur arises [5,6]. To prevent the decrease of the SNR and the range resolution owing to the distortion of echo caused by FDA, we propose an FDA compensation method. We generate an amplitude-modulated FM chirp pulse for transmission using FDA characteristics measured by transmitting a reference pulse and receiving the corresponding echo. By transmitting this chirp pulse, the echo is distorted by FDA, and, as a result, the echo is received as an ideal waveform.

Typical objectives of the amplitude modulation (AM) of an FM chirp signal are side-lobe suppression and the compensation of the resonance characteristics of a transducer [7]. The techniques for realizing those objectives are also integrated in our method. In this paper, we propose the

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additional AM of a transmitted signal for FDA compensation and confirm its effectiveness through experiments and simulations.

2. METHOD

2.1. Compensation of Transducer Characteristics

Before considering FDA, we should compensate the frequency characteristics of a transducer; this should be carried out for each transducer only once. To evaluate the frequency characteristics, an arbitrary FM chirp signal \( f(t) \) that should cover the frequency band to be used is applied to a transducer and the corresponding echo signal \( g(t) \) is observed. To avoid FDA, \( g(t) \) should be the echo from a reflector placed near the transducer. Using a frequency domain representation \( G(\omega) \) of \( g(t) \), we define the distortion function \( R_T(\omega) \) as follows:

\[
R_T(\omega) = H(\omega)/G(\omega),
\]

where \( H(\omega) \) is a frequency representation of an ideal FM chirp signal, which has a rectangular spectrum amplitude with a suitable window function. \( R_T(\omega) \) indicates a compensation coefficient of the transducer characteristics. Using \( R_T(\omega) \) and the frequency representation \( F(\omega) \) for \( f(t) \), we can adopt \( S(\omega) \) defined by the following equation as the FM chirp signal for which the transducer characteristics is compensated.

\[
S(\omega) = R_T(\omega)F(\omega).
\]

It is noted that this compensation is applied to both of the amplitude and the phase.

2.2. FDA Compensation Method

The proposed FDA compensation method is fundamentally the same as that for the transducer characteristics, but the FDA compensation should be performed for each depth that is desired to be finely imaged. To estimate the distortion caused by FDA in a propagation medium, \( s(t) \), which is a temporal representation of \( S(\omega) \) defined by Eq. (2), is transmitted focusing on the depth of interest. We refer to the echo signal as \( k(t) \). It can be assumed that FDA affects only the amplitude of the echo signal and not the phase. Therefore, the absolute value of \( R_{\text{FDA}}(\omega) \) in the next equation should be used as the FDA compensation coefficient.

\[
R_{\text{FDA}}(\omega) = H(\omega)/K(\omega),
\]

where, \( K(\omega) \) corresponds to \( k(t) \). The dashed line in Fig. 1 is an example of \( R_{\text{FDA}}(\omega) \) obtained by simulation mentioned in Sect. 4.3. We know that \( R_{\text{FDA}}(\omega) \) is distorted at the high-frequency part and there are several small ripple patterns, which may indicate the frequency characteristics of the propagation medium. They do not reflect the FDA characteristics and should not be compensated. Hence, in an actual procedure, we should approximate \( R_{\text{FDA}}(\omega) \) by function fitting as \( |R_{\text{FDA}}(\omega)|^{-1} \). The transmitted signal with FDA compensation can be formed as \( S_{\text{cmp}}(\omega) = |R_{\text{FDA}}(\omega)|^{-1} S(\omega) \). The amplitude-modulated FM chirp pulse corresponding to \( S_{\text{cmp}}(\omega) \) is transmitted, and its echo, which is expected to have no distortion caused by FDA, is used for imaging. It is noted that if the echo distorted by FDA is amplified using \( |R_{\text{FDA}}(\omega)|^{-1} \) instead of modifying the transmitted signal, the SNR is drastically lowered, particularly at the high-frequency part, and therefore, the imaging quality becomes low.

2.3. Determination of Compensation Coefficient

We study two approximation methods of \( |R_{\text{FDA}}(\omega)| \). In the first, we take the logarithm of \( |R_{\text{FDA}}(\omega)| \) and determine \( |R_{\text{FDA}}(\omega)|^{-1} \) as a linear approximation by line fitting, as shown in Fig. 1(a). In the following section, we explain the results of simulation using PZFlex, a standard finite-element method (FEM) code for an ultrasound propagation analysis. The definition of FDA in PZFlex is expressed as

\[
FDA = d(f_c/f_0)^n \quad [\text{dB/cm}],
\]

where \( d \) is the attenuation coefficient in a living body, \( f_c \) is the center frequency, \( f_0 \) is the measured frequency, and \( n \) is an exponent of FDA. Through the simulations assuming tissue for which FDA is suitably modeled by logarithmic linear characteristics, namely, \( n = 1 \), we confirmed that this method is effective [8]. However, a living body generally has logarithmic nonlinear characteristics. Therefore, it is thought that the linear approximation method cannot compensate FDA completely. Instead, we propose a curve-fitting compensation scheme in which logarithmic nonlinearity is assumed and adopt an arbitrary curve fitting using a fifth-order polynomial function to determine \( |R_{\text{FDA}}(\omega)| \), as shown in Fig. 1(b). To determine the order of the polynomial function, we actually performed polynomial curve fitting from the third to seventh order. As a result, it was confirmed that the half-width of the compressed echo envelope is essentially unchanged beyond the
fifth order. Therefore, in this study, we adopt the fifth-order polynomial function. In actual use, we should define and use a suitable determination criterion for the order. In the following experiments and simulations, we compare the performances of the two techniques.

3. EXPERIMENTS

3.1. Experiment Setup

To confirm the effectiveness of the proposed FDA compensation method, simple experiments were conducted. Figure 2 describes the experimental system. The transducer is SONIX IS1506R with a center frequency of 15 MHz. The amplifier is an Amplifier Research 50A15, the function generator is a Tektronix APG3102, and the oscilloscope is an IWATSU DS-5552. Signals were transmitted toward an iron plate placed 15 cm away from the transducer. We used a linear chirp of 8–12 MHz as a transmitted signal. The performance of the above-mentioned transducer is sufficient for transmitting and receiving the 8–12 MHz band. We observed echoes from the iron plate. Table 1 shows the measurement conditions.

3.2. Compensation of Transducer Characteristics

As a pre-adjustment, we performed the compensation of the transducer characteristics. Figure 3 indicates the evaluation results of the transducer characteristics. An ideal FM chirp signal with a Hanning window, as shown in Fig. 3(a), was applied to the transducer. The center frequency, the bandwidth and the duration in this evaluation are 10 MHz, 4 MHz, and 10 μs respectively. On the basis of the acoustic properties of water, there is very little absorption attenuation and diffusion attenuation. Therefore, when the propagation distance is short, FDA due to water can be ignored. After transmitting the pulse excited by the signal in Fig. 3(a), the distorted echo from the boundary between water and an iron plate placed 15 mm away from the transducer was received, as shown in Fig. 3(b). Using this echo signal and evaluating $R_T(\omega)$ in Eq. (1), we computed the signal whose echo is not affected by the transducer characteristics and show it in Fig. 4(a). We transmitted the pulse driven by the signal and received the echo signal shown in Fig. 4(b), which approximately equals the ideal signal. By comparing Fig. 3(c) and Fig. 4(c), we can confirm that the transducer characteristics is completely compensated. From this result, it is confirmed that the distortion caused by the transducer characteristics is completely removed.

3.3. Experimental Evaluation of FDA Compensation

In this section, we show the experimental results of the compensation method for FDA caused by the long propagation distance in water.

At first, Fig. 5(a) shows the echo signal obtained from the iron plate in water without FDA compensation. Figure 5(b) shows the power spectrum of the echo signal in (a). Figure 6 shows the distortion function $|R_{\text{FDA}}|$ obtained by experiment and $|R_{\text{FDA}}|^\text{app}$ that approximates $|R_{\text{FDA}}|$. Figure 6(a) shows the approximation by linear fitting, namely, linear compensation. Figure 6(b) shows the approximation by arbitrary curve fitting, namely, curve fitting compensation. Figure 5(c) shows the echo signal obtained from the iron plate with the linear compensation method applied once, and Fig. 5(e) shows the echo signal obtained from the iron plate with the linear compensation method applied once, and Fig. 5(e) shows the echo signal
obtained from the iron plate with the curve fitting compensation method applied once. Figures 5(d) and 5(f) show the power spectra of the echo signals in (c) and (e), respectively.

The normalized envelopes of the compressed echo signals before (Fig. 5(a)) and after linear compensation (Fig. 5(c)) are shown in Fig. 7(a) by dashed and solid lines, respectively. Similarly, Fig. 7(b) shows the effectiveness of the curve fitting compensation shown in Fig. 5(e) (solid line). Table 2 shows half-widths of the normalized envelopes of the compressed echo signals before and after compensation. From these results, we confirmed that the proposed compensation method is effective. In particular, curve fitting compensation seems to be superior to linear compensation.

4. SIMULATIONS

4.1. Objectives of Simulations

In future studies, it is necessary to evaluate the effectiveness of this method under conditions closer to those of a living body. Such evaluations are difficult by only experiments and simulations are required. Therefore, in this study, we examine the appropriateness of the simulation under the same conditions as those of the experiments described above. We examine the effectiveness of our methods through FEM simulations using PZFlex, in which the FDA characteristics obtained by the experiments are used for attenuation modeling. Additionally, we aim to confirm the necessity of repeated and recursive application of our FDA compensation methods.

4.2. Compensation of Transducer Characteristics

Similar to the experiments, we simulated the compensation for the transducer characteristics using the model shown in Fig. 8. In this model, a phased array transducer having a backing layer and a matching layer and consisting of 64 PZT elements is placed in a water tank and the FDA coefficient of water is set as 0.0 dB/cm/MHz. The left side
in Fig. 8 shows the backing layer, the 64 PZT elements are shown in blue, the center region shows water, and the right side shows iron. Beam forming was carried out for both transmission and reception. Figure 9 indicates the evaluation result of the transducer characteristics. An ideal FM chirp signal with a Hanning window, shown in Fig. 9(a), was applied to the transducer shown in Fig. 8. The center frequency, the bandwidth and the duration in this evaluation are 10 MHz, 4 MHz, and 5 μs, respectively. After transmitting the pulse excited by the signal shown in Fig. 9(a), the echo distorted by the characteristics of the transducer, for example, the influence of the frequency characteristics of the backing layer and the matching layer and diffusion attenuation due to a finite aperture, from the boundary between water and the iron plate was received, as shown in Fig. 9(b). Using this echo signal as \( g(t) \) and evaluating \( R_T(\omega) \) in Eq. (1), we computed the signal using Eq. (2), compensating the distortion by the transducer, and show it in Fig. 10(a). We computed again the transmission and the propagation of the pulse by applying the signal in Fig. 10(a) to the transducer and received the echo signal shown in Fig. 10(b), which approximately equals the ideal signal.

### 4.3 Simulation of FDA Compensation

Figure 11 describes the 2-dimensional simulation model including a transducer, a propagation medium and an imaging target. Transmitted pulses are formed using a linear array transducer model with 64 PZT elements, which is put at the left end of the medium shown in Fig. 11, and focused on a spot 30 mm away from the transducer. The medium shown in Fig. 11, consists of scatterers with which the speckle patterns imitating those of liver can be generated. The sound speed, density, and attenuation coefficient for each scatterers are randomly defined in the range shown in Table 3. The blue object shown in Fig. 11 imitates the tumor, and its characteristic is also shown in Table 3. We compute the echoes reflected from the above-mentioned regions, where the forward part of the target object exists, and analyze them. For receiving the echoes, similarly to transmission, beam forming is performed with the same focal point. Table 3 shows the sound speed, density and attenuation coefficient. The conditions of the transmitted signal are shown in Table 4. The slope of the FDA function, which is observed in the experiments using a phantom imitating a living body, is highly consistent with the slope of the attenuation function simulated with \( n = \ldots \).
Therefore, we set $n = 1.6$ in Eq. (4). Therefore, we set $n = 1.6$ in the following simulations. This means that FDA [dB] is varied nonlinearly with respect to the frequency as a tissue property.

The simulation results of the compensation method for FDA caused by a medium imitating liver are explained. The attenuation coefficient of liver is generally 0.6–1.0 dB, and we adopt 0.75–0.95 dB.

Figure 12(a) shows the echo signal without FDA compensation. Figure 12(b) shows the spectrum amplitude of (a). (c) echo signal with linear compensation, (d) spectrum amplitude of (c), (e) echo signal with curve fitting compensation, and (f) spectrum amplitude of (e).

Fig. 12 Simulation results of FDA compensation: (a) echo signal without FDA compensation, (b) spectrum amplitude of (a), (c) echo signal with linear compensation, (d) spectrum amplitude of (c), (e) echo signal with curve fitting compensation, and (f) spectrum amplitude of (e).

Fig. 13 Distortion function $|R_{FDA}|$ obtained by simulation (dashed line) and that approximated by (a) linear fitting (solid line) and (b) arbitrary curve fitting (solid line).

Fig. 14 Simulation results of compressed echo envelope: (a) compressed signal before compensation (dashed line) and after linear compensation (solid line), (b) compressed signal before compensation (dashed line) and after curve fitting compensation (solid line).

In the experiments and the simulations mentioned above, the proposed FDA compensation method is applied only once. Since $|R_{FDA}(\omega)|$ is an approximation of $|R_{FDA}(\omega)|$, there is a possibility of improving the compensation performance by recursively applying the method. Namely, at each repetition, we recalculate $|R_{FDA}(\omega)|$ and regenerate the transmitted signal by multiplying the previous transmitted signal by the recalculated $|R_{FDA}(\omega)|$ in the frequency domain. Therefore, we evaluated the half-width of the compressed echo signal with respect to the number of iterations of compensation. The results are shown in Fig. 15. For the linear compensation (dashed line), recursive application is more effective than a single application. On the other hand, the curve fitting compensation (solid line) yields good results without recursive application. However, too much repetition will complicate the attenuation shape in the frequency domain (Fig. 16), and therefore, the half-width of the compressed signal will decrease.

### Table 4 Conditions of transmitted signal.

| Condition                              | Value       |
|----------------------------------------|-------------|
| Frequency of transmitted signal [MHz]  | 10          |
| Band-width of frequency [MHz]          | 4           |
| Pulse-width [μs]                       | 5           |
| Window function of transmitted signal  | Hanning     |
| Focus of transmitted signal [mm]       | 30          |
5. CONCLUSIONS

In this study, we proposed a novel and simple FDA compensation method for fine imaging, and we confirmed that the curve fitting evaluation of FDA is effective. This method is based on the FM chirp pulse compression, and performs fine imaging effectively in a local manner by setting the depth of interest determined by pre-imaging by conventional B-mode imaging. Hereafter, we must evaluate the performance of our method experimentally using an actual human tissue. Additionally, the expansion to harmonic imaging is in progress.

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REFERENCES

[1] T. Misaridis and J. A. Jensen, “Use of modulated excitation signals in medical ultrasound,” IEEE Trans. Ultrason. Ferroelectr. Freq. Control, 52, 177–191 (2005).

[2] Z. Hu, T. Moriya and Y. Tanahashi, “Imaging system for intravascular ultrasonography using pulse compression technique,” Jpn. J. Appl. Phys., 40, 3896–3899 (2001).

[3] R. Y. Chiao and X. Hao, “Coded excitation for diagnostic ultrasound: a system developer’s perspective,” IEEE Trans. Ultrason. Ferroelectr. Freq. Control, 52, 160–170 (2005).

[4] I. Akiyama, N. Yoshizumi, S. Saito, Y. Wada, D. Koyama and K. Nakamura, “Development of multiple-frequency ultrasonic imaging system using multiple resonance piezoelectric transducer,” Jpn. J. Appl. Phys., 51, 07GF02-1-9 (2012).

[5] T. L. Szabo, Diagnostic Ultrasound Imaging: Inside Out (Elsevier, Amsterdam, 2004), Chap. 4.

[6] I. Akiyama, “On the effects of frequency dependent attenuation of biological tissues in ultrasonic imaging: Transmitting waveform relevant to imaging depth,” IEICE Tech. Rep., 111, pp. 43–48 (2012) (in Japanese).

[7] N. Levanon and E. Mozeson, Radar Signals (Wiley-Interscience, Hoboken, N.J., 2004).

[8] K. Koumoto, N. Tagawa, K. Okubo and I. Akiyama, “Wide band pulse compression imaging with transmission compensation for frequency dependent attenuation,” IEEE Int. Ultrason. Symp., pp. 1658–1661 (2012).