Published in final edited form as:

IEEE Trans Ultrason Ferroelectr Freq Control. 2008 February ; 55(2): 508–513. doi:10.1109/TUFFC.

2008.670.

Chirp coded excitation imaging with a high-frequency ultrasound annular array

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Abstract

High-frequency ultrasound (HFU, > 15 MHz) is an effective means of obtaining fine-resolution images of biological tissues for applications such as ophthalmologic, dermatological, and small-animal imaging. HFU has two inherent drawbacks. First, HFU images have a limited depth of field (DOF) because of the short wavelength and the low fixed F-number of conventional HFU transducers. Second, HFU can be used only to image a few millimeters deep into a tissue because attenuation increases with frequency. In this study, a five-element annular array was used in conjunction with a synthetic-focusing algorithm to extend the DOF. The annular array had an aperture of 10 mm, a focal length of 31 mm, and a center frequency of 17 MHz. To increase penetration depth, 8- μ s, chirp-coded signals were designed, input into an arbitrary waveform generator, and used to excite each array element. After data acquisition, the received signals were linearly filtered to restore axial resolution and increase the SNR. To compare the chirp-coded imaging method with conventional impulse imaging in terms of resolution, a 25- μ m diameter wire was scanned and the -6 dB axial and lateral resolutions were computed at depths ranging from 20.5 to 40.5 mm. The results demonstrated that chirp-coded excitation did not degrade axial or lateral resolution. A tissue-mimicking phantom containing 10- μ m glass beads was scanned, and backscattered signals were analyzed to evaluate SNR and penetration depth. Finally, *ex vivo* ophthalmic images were formed and chirp-coded images showed features that were not visible in conventional impulse images.

I. INTRODUCTION

High-frequency ultrasound (HFU, > 15 MHz) offers an exciting potential for noninvasive visualization of living tissues at the microscopic level. Ultrasound is non-ionizing, non-invasive, real-time, and an order of magnitude less expensive than other common imaging modalities (e.g., magnetic resonance imaging). However, the fine-resolution advantages offered by HFU are offset by limitations in penetration depth caused by frequency-dependent attenuation and limitations in depth-of-field (DOF) when low F-number transducers are employed to improve cross-range resolution. Attenuation of ultrasound in tissue increases with frequency [1], [2] and, therefore, current uses of HFU are limited to applications that do not require deep penetration to image the tissue of interest [3]–[5].

This study aims to significantly improve HFU image quality by using two independent approaches. The first approach uses synthetic-focused annular arrays with overall apertures similar to typical spherically-focused transducers to increase DOF [6], [7]. The radial symmetry of annular arrays leads to a high-quality radiation pattern while employing fewer elements than linear or phased arrays. However, annular arrays need to be mechanically scanned to obtain a 2D image. We have established that annular arrays increase DOF [6], [7] and, therefore, the present paper only briefly reviews annular array fabrication and synthetic focusing methods.

The second HFU imaging approach uses coded excitations (i.e., engineered excitation pulses) that are capable of increasing the effective penetration depth by improving signal-to-noise ratio (SNR) [8], [9]. Resolution and penetration depth are critically important for medical ultrasound imaging. Normally, these two properties present a tradeoff, in which one property can be improved only at the expense of the other. However, Chiao has demonstrated that coded excitation is capable of extending the limit associated with this tradeoff [9]. Coded excitation permits the SNR to be increased through appropriate encoding on transmit and decoding on receive. In this study linear-chirp signals were used to excite the annular-array transducer. The objectives of this study are to demonstrate that chirp annular-array imaging can lead to better image quality than current state-of-the-art HFU images. The described methods are general and are applicable to a vast range of clinical applications, including, ophthalmological, dermatological, and gastrointestinal imaging.

II. METHODS

A. Annular Array and synthetic focusing

The annular array used in these studies was fabricated using previously described methods [6]. A 25- μm thick poly(vinylidene fluoride-tetrafluoroethylene), P(VDF-TrFE), membrane (Ktech Corp., Albuquerque, NM) with one side metallized was bonded to a copper-clad polyimide film with non-conductive epoxy. The final array consisted of five, equal-area annuli with a 10-mm total aperture and a 31-mm geometric focus. The spacing between the annuli was 100 μm . Table I summarizes the key performance parameters (center frequency (f_c), relative amplitude measured at f_c (dB@ f_c), 6 dB-bandwidth (BW_{6dB}), insertion loss (IL) measured at f_{IL}) for the five array elements after impedance matching (3.3 μH). Measurements were made as previously described [6].

A synthetic-focusing algorithm was developed to permit the focus of the annular array to be axially shifted by post-processing the acquired data [7]. To focus the array to a depth d on transmit, the time delay t_n required for ring n of the array is $t_n = [a_n^2(1/R - 1/d)]/2c$ where R is the geometric focus (i.e., 31 mm) and a_n is the average radius of ring n [10]. The round trip delay is then the sum of the transmit and receive delays giving $t^{tot} = t^T + t^R$. To focus at a depth d , t^{tot} is calculated for all 25 transmit/receive pairs. The delays are then applied to the A-lines and the resulting signals are summed. In order to increase the DOF over a fixed axial span, d is shifted in intervals of Δd . Then, an overall image is formed by windowing the data at each focal depth and assembling the windowed data into a final composite image. In all of our experiments, Δd is chosen to be 0.4 mm. (If $t^{tot} = 0$ is utilized for all of the delays, then the results simulate a single-element transducer with a total aperture of 10 mm and a geometric focus of 31 mm.)

B. Chirp Coded Excitation

In this study, only linear, chirp-coded excitations were used. A linear chirp is a coded signal that linearly spans a frequency bandwidth $B = f_2 - f_1$, where f_1 and f_2 are the starting and ending frequencies, respectively. If the chirp sweeps from f_1 to f_2 over a time, T , then the chirp coded excitation is described by

$$s(t) = w(t) \cos(2\pi f_1 t + \pi b t^2), \quad (1)$$

where $w(t)$ is a windowing function that vanishes outside of $t \in [0, T]$ and b is the sweep rate equal to $(f_2 - f_1)/T$. T is usually an order of magnitude longer than the impulse response of the transducer. In this study, T was set to 8 μs for the 17-MHz annular array and decoding was accomplished by linearly filtering the backscattered signals. The linear filter is often called a

compression filter because it is meant to restore axial resolution [9]. The compression filter used in this study consisted of the time-reversed coded chirp (i.e., matched filter) weighted by a Chebyshev window (i.e., “mismatched filter”). Theory predicts that when there is no tapering on the chirp and a matched filter is used, the gain in SNR is equal to the time-bandwidth (TB) product [11]. Compression was performed before synthetic focusing to avoid artifacts due to the long-time duration of the chirp with respect to the DOF at the geometric focus of the transducer [8]. The counterpart of doing the compression before the focusing is that 25 compressions (i.e., one for each transmit-to-receive combination) are necessary instead of one (i.e., for the synthetically-focused signal). Compressing first would become impractical with arrays with a much greater number of elements (e.g., > 15) and shorter chirps would need to be used [8].

Eq. (1) indicates that there are four primary chirp-design parameters: f_1, f_2, T , and $w(t)$. We designed two chirps (termed C1 and C2, Figs. 1a and 1c) based on the bandwidths for each array element from quartz-plate reflections (Table II); these values were consistent with values derived in theoretical simulations used to predict improvements in SNR (i.e., $TB > 20$ dB).

(We define an $x\%$ -Tukey window as a window such that $w(t) = 1$ when $t \in \left[\frac{x}{100}T, \frac{1-x}{100}T \right]$, such that $w(t)$ is the first (i.e., rising) half of a Tukey window when $w(t) < \frac{x}{100}T$ and such that $w(t)$ is the second (i.e., falling) half of a Tukey window when $w(t) > \frac{1-x}{100}T$ [12]).

In all of our experiments, the compression filter consisted of the time-reversed excitation chirp (i.e., C1 or C2) weighted by a Chebyshev window with a prescribed sidelobe level of -80 dB. The compression filters matched to C1 and C2 are termed MF1 and MF2 (Figs. 1b and d).

Figure 1e displays the normalized spectra of C1, C2, MF1, and MF2. The absence of Fresnel ripples in the spectra for C1 and C2 resulted from the use of a Tukey window. Figure 1e also displays the spectrum of the simulated impulse response of a transducer with a center frequency of 17 MHz and a fractional bandwidth of 30%. These values are very similar to the measured properties of the array transducer (Table I). To appreciate how coded excitation can increase SNR, white noise was added to the simulated impulse response. The resulting simulated impulse response had an SNR of 45 dB, which is in the range of most ultrasound imaging systems. Finally, Figure 1f shows the log-compressed envelope of the two compression outputs and of the simulated noisy impulse response. These results reveal that the impulse response had the smallest main lobe width, and thus the best axial resolution (Table III). However, the chirp excitations led to an increase in SNR of > 14 dB (Table III).

C. Image Acquisition

Image data were acquired with the annular array by making five scan passes across the test object. On each pass, one of the array elements was excited and the receive echos were digitized at 200 MHz on all five channels. In this way, we were able to acquire the full set of 25 transmit-to-receive combinations. After saving the data, each digitized signal was linearly filtered and compressed with a mismatch filter. The compressed signals were delayed and summed for synthetic focusing [7], envelope-detected, and log-compressed for B-mode display. We also acquired data using a mono-cycle pulser (Avtech AVB2-TA-C-CVA, Ottawa, Ontario, Canada) in order to make a comparison to the “gold standard”. The scanning method and image formation for the mono-cycle excitation were the same as for the chirp imaging method except that no compression was performed.

The basic experimental setup was described in detail in a previous publication for impulse imaging [7]. For chirp-coded excitation imaging, the pulser-receiver unit was replaced by an

arbitrary waveform generator (WW1281, Tabor Electronics, Haifa, Israel) followed by a linear broadband amplifier with a gain of 50 dB (ENI 350L, Rochester, NY). To design and input coded waveforms into the arbitrary waveform generator, custom software was designed in Matlab (The MathWorks, Natick, MA) and LabVIEW (National Instruments, Austin, TX).

D. Experiments

We performed three sets of experiments to evaluate the effectiveness of chirp imaging versus standard impulse imaging. Our first experiment compared the -6 -dB axial and lateral resolutions of conventional impulse and chirp-compression imaging. It consisted of scanning across $25\text{-}\mu\text{m}$ diameter tungsten wires placed at depths ranging from 20.5 mm to 40.5 mm. At each depth, two images were formed, one using coded excitation and one using impulse excitation. In these experiments, C1 and MF1 were used as the excitation chirp and the compression filter. The final images were also synthetically focused. Our second experiment investigated whether coded excitation increases SNR and penetration depth relative to what can be achieved with conventional-impulse imaging. For this experiment, C1 and MF1 were used and a tissue-mimicking phantom (ATS Laboratories Bridgeport, CT) containing $10\text{-}\mu\text{m}$ diameter glass beads (8×10^6 beads/cm³) was scanned using chirp and impulse excitation. The phantom had an attenuation of 0.5 dB/cm/MHz near 6 MHz. Images were formed for each imaging method, with and without synthetic focusing. Finally, we evaluated the chirp imaging method *in vitro*. Images of a bovine eye (using C2 and MF2) and human eye-bank eye (using C1 and MF1) were formed for each imaging method, with and without synthetic focusing. The images were then qualitatively assessed.

III. RESULTS

A. Axial and Lateral Resolutions

Figure 2a displays the -6 -dB axial resolution (in units of wavelength at 17 MHz) as a function of depth for chirp and impulse imaging. The results demonstrate that both techniques yield axial resolutions within 5% of each other, except at the depth of 21.5 mm where the chirp axial resolution is 10% better than the impulse axial resolution. Figure 2a also shows that the axial resolution is the best near the geometric focus of the transducer (i.e., 31 mm) for both imaging methods. However, synthetic focusing significantly increased the DOF as evidenced by the fact that axial resolution remains below 1.6λ over depths ranging from 26 to 36 mm.

Figure 2b displays the -6 -dB lateral resolution as a function of depth for both imaging techniques. Figure 2b demonstrates that the lateral resolution achieved by both techniques was essentially the same (as theory predicts). These figures show that the lateral resolution worsens (i.e., increases) with depth because, as expected, the synthetic focusing algorithm increases the effective transducer F-number as the focal depth increases with a fixed aperture.

B. SNR and Penetration Studies

Figure 3 shows the phantom images. The dynamic range of these images was set to 60 dB to help appreciate how the methods perform in terms of SNR. For both imaging techniques, synthetic focusing permitted deeper penetration into the phantom as evidenced by the backscatter from the glass beads. Also, coded excitation led to an increase in SNR independently of whether the synthetic focusing algorithm was used (illustrated by the darker background colors of Figs. 3b and 3d compared to those of Figs. 3a and 3c).

To quantify the above observations, we examined the average backscatter from each of the four images as a function of depth (Fig. 3e). In Figure 3e, the four curves clearly show a peak at a depth of 22 mm corresponding to the front of the scattering phantom. After the peak, the four curves indicate that the average backscatter decreased with depth and then plateaued. We

used this plateau as a basis for defining SNR. Table IV displays the SNRs for each of the four imaging methods. The SNR can be interpreted as the absolute average magnitude of the noise plateau because the magnitude of the front echo of the phantom was normalized to 0 dB. As we observed qualitatively in the images of Fig. 3, the SNR increased with dynamic focusing and coded excitation. In particular, an SNR increase of 11.9 dB was observed between the synthetically-focused chirp and impulse excitation. The last column of Table IV presents the penetration depth into the phantom. These values (symbolized in Figure 3e by the cross symbol) demonstrate that synthetic focusing and chirp excitation contribute individually to increasing the effective penetration depth. The maximum effective penetration depth (37.5 mm, i.e., ≈ 15.5 mm into the phantom) was obtained when both methods were combined.

C. Ophthalmic In Vitro Imaging

Three ultrasound images of a bovine eye are displayed in Figure 4. In all three images, the geometric focus of the transducer (31 mm) was just beyond the lens. Figure 4a can be interpreted as the current state of the art for ultrasound ophthalmologic imaging because with impulse excitation and without synthetic focusing, the annular array is equivalent to a single-element transducer. The anterior segment of the eye was in the defocused near-field in images formed without synthetic focusing and, thus, this area was poorly resolved (Fig. 4a).

As shown in Fig. 4b, synthetic focusing provided a dramatic improvement in image quality. This figure clearly showed that the anterior and posterior segments were in sharp focus. A detachment of the corneal epithelium, which was not observed in Fig. 4a, was clearly depicted in the synthetic-focus image (Fig. 4b). Similarly, the anterior chamber was more clearly visualized with synthetic focusing. Also the SNR increased by 5 dB compared to the unfocused case.

Further improvement in image quality was observed in the chirp image (Fig. 4c); not only was the whole eye in sharp focus, but the increase in SNR (+17 dB compared with the focused impulse image) allowed visualization of structures that previously were hidden by noise. The lens circumference was fully visible, and scattering from the vitreous was seen proximal to the retina. Figure 4c demonstrates the strength of combining annular array imaging with coded excitation. The annular array allows for the creation of an image with an extended DOF and the chirp allows for deeper penetration and for the display of weaker reflectors. The practical implication is that annular array imaging with synthetic focusing and chirp excitation could be diagnostically significant.

IV. DISCUSSION AND CONCLUSIONS

The initial objectives of this study were successfully achieved because we were able to generate images of better quality than current state-of-the-art HFU scanners by combining coded-chirp signals with annular arrays. We demonstrated that axial and lateral resolutions were not degraded, and that we could image deeper using chirps. We also demonstrated with an *in vitro* example, that our synthetically focused chirp-derived images contained additional diagnostically-significant information for detecting and assessing ocular pathologies when compared to current state of the art methods. A number of disease conditions of the eye involve the visualization of relatively faintly reflecting or scattering structures within the vitreous. These conditions would be more likely detected with the improved SNR from chirp-based imaging and the improved DOF from synthetically focusing the annular array.

Overall, synthetically focused annular arrays with chirp-based imaging could be easily implemented easily in clinical scanners to produce higher-quality images than clinicians currently utilize because the post-processing simply consists of linearly filtering all the signals

with a filter that has a well defined impulse response. Linear filtering can be performed directly on a DSP chip or off-line by replacing a convolution by a multiplication in the Fourier domain.

In conclusion, chirp coded-excitation represents a valuable method for improving HFU images. Implementation in clinical instruments could be done in a manner that is transparent to the end user. The chirp images would have the same format and properties as familiar, conventional, B-mode images except for a noticeably improved SNR. The salient advantage of such images is that their improved quality (in terms of DOF and SNR) can help detection and evaluation of disease conditions when visualization of fine tissue structures necessary. Once the chirp imaging system is optimized, it will be able to produce higher-quality images than possible with conventional systems. Finally, because chirps seem better adapted than other coded signals to HFU imaging, chirp imaging has the potential to become an important method for generating high-quality HFU images in research and clinical applications.

ACKNOWLEDGMENT

This research was supported by NIH grants EB006509, EY014371, and the Riverside Research Institute Fund for Biomedical Engineering Research.

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Fig. 1.

a)–d) Plots of the two chirp signals and their corresponding compression filters. e) Spectra of the two chirps, the two compression filters, and the simulated impulse response of a 17-MHz center-frequency transducer with a 30% fractional bandwidth and SNR of 45 dB. f) Envelope-detected and log-compressed display of the simulated noisy impulse response and the two compression outputs.

Fig. 2. Axial (a) and lateral (b) -6-dB resolutions obtained as a function of depth using chirp and impulse excitation.

Fig. 3.

Images of a tissue mimicking phantom containing 10- μ m diameter glass beads using different imaging methods: a) impulse imaging without synthetic focusing, b) impulse imaging with synthetic focusing, c) chirp imaging without synthetic focusing, and d) chirp imaging with synthetic focusing. C1 and MF1 were used in this experiment.

Fig. 4.

Ex vivo images of a bovine eye: a) impulse without synthetic focusing, 60-dB dynamic range, b) impulse with synthetic focusing, 60-dB dynamic range, and c) chirp with synthetic focusing, 75-dB dynamic range. C2 and MF2 were used in this experiment.

TABLE I

TRANSDUCER PERFORMANCE SUMMARY

Ring	f_c (MHz)	dB@ f_c	BW _{6dB} (%)	BW _{6dB} (MHz)	IL (dB)	f_{IL} (MHz)
1	17	0	28	4.8	17	18
2	17	-9	25	4.3	25	17
3	17	-12	34	5.8	28	17
4	17	-13	35	6.0	31	17
5	17	-13	31	5.3	30	17

TABLE II
PARAMETERS OF CHIRP SIGNALS USED IN THE EXPERIMENTS.

Name	f_1 (MHz)	f_2 (MHz)	T (μ s)	B (MHz)	TB (dB)	w
C1	6.5	32	8	25.5	24.0	9%-Tukey
C2	0	40	8	40	25.1	15%-Tukey

TABLE III
AXIAL RESOLUTION AND SNR COMPARISON FROM SIMULATED IMPULSE RESPONSE

Name	−6 dB Axial resolution ($\lambda_{17\text{ MHz}}$)	SNR (dB)	SNR improvement (dB)
Impulse	1.12	44.6	NA
C1	1.34 (19% worse)	59.8	15.2
C2	1.22 (7.5 % worse)	59.1	14.5

TABLE IV**SNR AND PENETRATION DEPTH RESULTS**

Method	Dynamic Focusing	SNR (dB)	Penetration depth(mm)
Impulse	No	30.3	29.1
Impulse	Yes	46.9	33.8
Chirp	No	39.4	31.6
Chirp	Yes	58.8	37.5