Published in final edited form as:

Ultrasonics. 2013 February ; 53(2): . doi:10.1016/j.ultras.2012.07.003.

Combined chirp coded tissue harmonic and fundamental ultrasound imaging for intravascular ultrasound: 20–60 MHz phantom and *ex vivo* results

Jinhyoung Park*, Xiang Li¹, Qifa Zhou², and K. Kirk Shung³

NIH Resource on Medical Ultrasonic Transducer Technology, Department of Biomedical Engineering, University of Southern California, Los Angeles, CA 90089, USA

Abstract

The application of chirp coded excitation to pulse inversion tissue harmonic imaging can increase signal to noise ratio. On the other hand, the elevation of range side lobe level, caused by leakages of the fundamental signal, has been problematic in mechanical scanners which are still the most prevalent in high frequency intravascular ultrasound imaging. Fundamental chirp coded excitation imaging can achieve range side lobe levels lower than –60 dB with Hanning window, but it yields higher side lobes level than pulse inversion chirp coded tissue harmonic imaging (PI-CTHI). Therefore, in this paper a combined pulse inversion chirp coded tissue harmonic and fundamental imaging mode (CPI-CTHI) is proposed to retain the advantages of both chirp coded harmonic and fundamental imaging modes by demonstrating 20–60 MHz phantom and *ex vivo* results. A simulation study shows that the range side lobe level of CPI-CTHI is 16 dB lower than PI-CTHI, assuming that the transducer translates incident positions by 50 μm when two beamlines of pulse inversion pair are acquired. CPI-CTHI is implemented for a proto-typed intravascular ultrasound scanner capable of combined data acquisition in real-time. A wire phantom study shows that CPI-CTHI has a 12 dB lower range side lobe level and a 7 dB higher echo signal to noise ratio than PI-CTHI, while the lateral resolution and side lobe level are 50 μm finer and –3 dB less than fundamental chirp coded excitation imaging respectively. *Ex vivo* scanning of a rabbit trachea demonstrates that CPI-CTHI is capable of visualizing blood vessels as small as 200 μm in diameter with 6 dB better tissue contrast than either PI-CTHI or fundamental chirp coded excitation imaging. These results clearly indicate that CPI-CTHI may enhance tissue contrast with less range side lobe level than PI-CTHI.

Keywords

Chirp coded tissue harmonic imaging; Chirp coded excitation; Intravascular ultrasound imaging

© 2012 Elsevier B.V. All rights reserved.

* Corresponding author. Address: Department of Biomedical Engineering, University of Southern California, 1042 Downey Way, University Park, DRB132, USA. Tel.: +1 213 821 2647; fax: +1 213 821 3897. jinhyoung.park@gmail.com (J. Park). xiangli@usc.edu (X. Li), qifazhou@usc.edu (Q. Zhou), kkshung@usc.edu (K.K. Shung).

¹Address: Department of Biomedical Engineering, University of Southern California, 1042 Downey Way, University Park, DRB132, USA. Tel.: +1 213 821 2647; fax: +1 213 821 3897.

²Address: Department of Biomedical Engineering, University of Southern California, 1042 Downey Way, University Park, DRB162, USA. Tel.: +1 213 740 9475; fax: +1 213 821 3897.

³Address: Department of Biomedical Engineering, University of Southern California, 1042 Downey Way, University Park, DRB135, USA. Tel.: +1 213 821 2653; fax: +1 213 821 3897.

1. Introduction

Tissue harmonic imaging (THI) is known to yield lower side lobe levels resulting in finer lateral resolution and tissue contrast than fundamental frequency imaging [1]. These advantageous characteristics allow subtle lesions in an organ to be delineated. Hohl et al. [2] found that THI offers better tissue contrast in pancreatic head regions than fundamental imaging. However, THI may suffer from a reduced dynamic range, usually at least 20 dB below the fundamental signal, and from degradation in spatial resolution caused by the increased attenuation at harmonic frequencies as well as limitations in bandwidth [3]. The reduced echo signal to noise ratio (eSNR) may be compensated by applying coded excitation techniques to THI. Chiao and Hao [4] demonstrated that pulse inversion chirp coded THI (PI-CTHI) improved the depth of penetration in imaging liver compared with short burst THI. The pulse inversion tissue harmonic imaging methods sum two beamlines acquired by transmitting a burst and its amplitude inverted alternately to cancel out signals on fundamental frequencies.

Recently, THI has been employed on high frequency (40 MHz) ultrasound imaging systems for intravascular applications. Goertz et al. [5] demonstrated that by applying THI to intravascular ultrasound (IVUS), tissue contrast was improved by 22 dB as compared to the fundamental imaging. However, THI for IVUS also suffered from the reduced eSNR. Some approaches, e.g. a dual frequency imaging [6], and signal averaging [7], were shown to be capable of enhancing eSNR. These methods, however, either require specially designed transducers or may cause degradation in lateral resolution.

The application of PI-CTHI to IVUS may increase tissue contrast and eSNR without a need to modify either the design of the transducer or image acquisition methods. In mechanical scanning IVUS, however, the transducer incident position is constantly changed. This operation causes discrepancies of echo waveforms between the non-inverted and inverted signal pair. It results in fundamental signal leakages [8] which may cause the elevation of range side lobe level (RSLL) in PI-CTHI. The use of fundamental chirp coded excitation imaging (F-CI) solely yields a low RSLL of -50 dB for a mechanical scanner [9], but the increased side lobe level and resultant artifacts degrade spatial resolution and cause blurred imaging presentation [10].

Previously, studies on combining harmonic imaging with fundamental imaging reported advantages of both harmonic and fundamental imaging. For example, image fusion approaches, which combine the separate frames of the harmonic and fundamental image, were shown to perform better presentation on tissue contrast and eSNR than images obtained from each approach alone [11,12]. In these studies image fusion was achieved offline. Nowicki et al. [13] presented dual tone coding which can transmit a combined multi-cycled sinusoidal burst having both fundamental and harmonic frequencies. In this study, however, the axial resolution is worse than the lateral resolution because of the multiple cycled two-tone burst which may not be acceptable for visualizing blood vessel structures smaller than a few hundred microns with 40 MHz IVUS.

In the present work, a new method of combined pulse inversion chirp coded tissue harmonic imaging (CPI-CTHI) is presented with a custom-built rotational mechanical scanner, a proto-typed IVUS capable of real time data acquisitions. A transmit waveform for CPI-CTHI is designed by combining the chirped burst for F-CI with a sub-harmonic burst for PI-CTHI. The echo can be compressed by a single application of convolution with a matched filter. Note that the matched filter is a time reversed fundamental chirp coded waveform. Experiments have been carried out on a wire phantom to assess the performance of CPI-CTHI and the results are compared with simulation data. *Ex vivo* images of rabbit trachea

acquired show improvements in visualizing blood vessels of small diameter (<200 μm) due to the lower RSL and higher eSNR than PI-CTHI.

2. Principles and implementation

The transmit waveform for CPI-CTHI is implemented by combining the two waveforms for PI-CTHI and F-CI. The concept of CPI-CTHI for IVUS is evaluated with a non-linear simulator, and carried out for a proto-typed IVUS.

2.1. Waveform generation

A fundamental chirp waveform is tapered with a Hanning window to reduce RSL and generated by the following equation [14]:

$$\text{Chirp_Fund}(t) = \cos \left[2\pi \left(f_i + \frac{\alpha t}{2} \right) t \right] \times W_{\text{Hanning}} \quad (1)$$

where Chirp_Fund is a fundamental chirp waveform, f_i is the lowest frequency of the chirp, α is the rate of linear increase in frequency, t is time and W_{Hanning} is Hanning window. In the experimental arrangement, the chirped burst sweeps from 20 MHz to 60 MHz linearly for 1 μs . The burst can be compressed by convolution with a matched filter which is a time reversed transmitted waveform.

The waveforms for PI-CTHI are designed by the following equation:

$$\text{Chirp_Harm_In}(t) = -\text{Chirp_Harm_Out}(t) = \cos \left[2\pi \left(f_i + \frac{\alpha t}{2} \right) \frac{t}{2} \right] \times \sqrt{W_{\text{Hanning}}} \quad (2)$$

where Chirp_Harm_In and Chirp_Harm_Out are a pulse inversion pair for PI-CTHI and other symbols are the same as in Eq. (1). To use the matched filter for PI-CTHI same with the one used for F-CI, frequencies are half of the fundamental waveform of Eq. (1), and window function is root-squared. Note that window functions used for PI-CTHI become squared in the echo on second harmonic frequencies [15].

The final transmission waveform for CPI-CTHI is formed by summing these two previous equations with weightings shown as follows.

$$\begin{aligned} \text{Chirp_Final_In}(t) &= (A) \times \text{Chirp_Fund}(t) + (B) \times \text{Chirp_Harm_In}(t) \\ \text{Chirp_Final_Out}(t) &= (A) \times \text{Chirp_Fund}(t) + (B) \times \text{Chirp_Harm_Out}(t) \\ \text{Weighting}(dB) &= 20 \times \log_{10} \left(\frac{B}{A} \right) \end{aligned} \quad (3)$$

where Chirp_Final_In and Chirp_Final_Out are a pulse inversion pair for CPI-CTHI and A and B are weighting values. Note that the weighting values are given in the decibel scale described by the third equation of (3). Fig. 1 describes Eq. (3), and a 'Weighting' of 0 dB yields the final transmit waveform of $\text{Chirp_Final_In}(t)$.

The 'Weighting' value is a controllable parameter which decides the ratio between harmonic and fundamental signals and is optimized through non-linear simulation. Once the combined bursts are administered to a target, both fundamental and harmonic frequencies from Chirp_Fund and Chirp_Harm_In (or Out) exist in the echoes. Summations of the scanlines acquired from the pulse inversion pair in Eq. (3) cancel Chirp_Harm_In (Out) and results in Eq. (4).

$$\begin{aligned}
 FinalEcho(t) &\approx A \cos \left[2\pi \left(f_i + \frac{\alpha t}{2} \right) t \right] \\
 &\times W_{Hanning} + B \cos \left[2\pi \left(f_i + \frac{\alpha t}{2} \right) t \right] \\
 &\times W_{Hanning} + E \cos \left[2\pi \left(f_i + \frac{\alpha t}{2} \right) \frac{t}{2} \right] \\
 &\times \sqrt{W_{Hanning}}
 \end{aligned} \quad (4)$$

where A is the amount of fundamental signal from *Chirp_Fund*, B is the harmonic signal from *Chirp_Harm_In (Out)* and E is the residue, not canceled out from the summation of the pulse inversion pair.

2.2. Numerical simulation of nonlinear field

The new imaging mode is simulated with an open source for analyzing non-linear acoustic propagation fields [16]. Harmonic distortion is analyzed in a 2D field using the equation known as the “ B/A ” model given as follows:

$$\rho \frac{\delta^2 u}{\delta t^2} = -\nabla p, \quad p = K \left(\nabla \cdot u + \frac{1}{2} \frac{B}{A} (\nabla \cdot u)^2 \right) \quad (5)$$

where p is acoustic pressure, ρ density in the propagating medium, K bulk modulus (ρc^2 , c sound speed), B/A nonlinearity parameter of the medium, and u particle displacement vector. The open-source code is modified to model an unfocused transducer having an aperture size of 0.5 mm with the center frequency of 35 MHz and the bandwidth of 75%. The simulated 2D field is divided with rectangular grids having size of 5.3 μm by 5.3 μm . The pressure values at the grids are updated by the Eq. (5) every 3.5 ns which is waveform propagation time in each grid. From the simulation, a point spread function (PSF) is calculated at 1.4 mm, natural focus of the simulated transducer. PSFs are acquired twice for the pulse inversion pair described by Eqs. (2) and (3) and summed up to cancel the fundamental frequencies from *Chirp_Harm_In (Out)*. The frequency ranges of Hanning windowed Chirped waveforms for the fundamental and the harmonic imaging are from 20 MHz to 60 MHz and from 10 MHz to 30 MHz respectively. Note that the use of Hanning window reduces the bandwidth of chirped waveform to half of the original. The effectiveness of ‘*Weighting*’ is investigated within the range from –20 dB to 20 dB. Simulation parameters are provided in Table 1.

To simulate the effects caused by motion artifacts, one of the two-dimensional PSFs among the pulse inversion pair is shifted in the lateral direction from 0 μm to 60 μm .

2.3. System configuration

A custom-designed rotational mechanical scanner moves a target around a fixed tip of a side-viewing high frequency needle transducer. The system setup is depicted in Fig. 2. A pulse generator transmits bursts at a 2 kHz pulse repetition rate to a single element transducer, and the received echo signals are pre-amplified by a pulser/receiver (5900PR, Olympus, MA, USA) and digitized with an analog to digital converter (CS12400, GaGe Applied Technologies, Inc., Lachine, QC, Canada) at 400 MHz sampling rate. Note that the scanning speed is adjusted to acquire 400 scanlines per one rotation. Particularly, the combination of a two-channel function generator (AFG3252, Tektronix, Beaverton, OR, USA) and a power amplifier (75A250A, Amplifier Research, Souderton, PA, USA) is employed for the generation of chirp coded excitation burst. The frequency ranges of

Hanning windowed Chirped waveforms for the fundamental and the harmonic imaging are from 20 MHz to 60 MHz and from 10 MHz to 30 MHz respectively. In combining the two waveforms (harmonic and fundamental pulses), the ‘*Weighting*’ values are varied from –20 dB to 20 dB. Particularly, a burst and its amplitude inverted counterpart are alternatively transmitted from the different channels of the function generator for the pulse inversion operation. The unfocused needle transducer was fabricated from PMN-PT (lead magnesium niobate-lead titanate) single crystal having an aperture of $0.5 \times 0.5 \text{ mm}^2$ with a center frequency of 38 MHz and –6 dB fractional bandwidth of 41%. Note that PNM-PT can be fabricated for small aperture size transducer due to the relatively high dielectric permittivity (~500) compared with the other single crystal materials, e.g., LiNbO_3 (~39).

The digitized signal is transferred to the backend software developed with LabView (LabView 5.0, National Instruments, Austin, TX, USA) for real time data acquisition. The echo data are acquired by five frames per second, and one frame of the saved data is processed for envelop detection by Hilbert transformation, log compressed and gray mapping with Matlab (Matlab R2010b, Mathworks, Natick, MA, USA) programming. A block diagram for the post-processing is depicted in Fig. 3.

3. Experimental measurement

To evaluate the performance of CPI-CTHI, wire phantom studies and *ex vivo* experiments have been carried out. Through wire phantom studies, eSNR and RSL are measured. *Ex vivo* imaging is performed to demonstrate the enhancement on tissue contrast of CPI-CTHI.

3.1. Wire phantom and sphere phantom

A wire phantom consisting of two wires was immersed in deionized (DI) water in a small water tank ($6.5 \times 6.5 \times 4.3 \text{ cm}$). The thin wires (20 μm diameter) were aligned in parallel with the transducer's needle shaft for scanning. The transducer was also immersed in the water tank. The distances from the transducer's surface to the target were adjusted to place the first wire at the natural focus of the transducer (2.9 mm). Note that the diagonal length (0.7 mm) of the transducer is considered as aperture size in calculating the natural focus. From the acquired wire target images, the eSNR was analyzed from the following equation [17] for which a noise frame was acquired at 0 V_{pp} input by disabling the function generator (AFG3252, Tektronix, Beaverton, OR, USA).

$$eSNR = 20 \times \log_{10} \left(\frac{\text{Maximum}(\text{abs}(\text{WireTargetFrame}))}{\sigma(\text{NoiseFrame})} \right) \quad (6)$$

where σ is the standard deviation and *abs* is the absolute value of a signal. *eSNR* is given in the decibel scale. The RSL, which is the level of the second highest lobe around the main lobe in the axial direction, in the wire target image using the harmonic chirp was measured for CPI-CTHI, and the result was compared with that acquired with PI-CTHI and F-CI.

3.2. Ex vivo experiment

A trachea sample was harvested from a healthy rabbit and immersed in DI water. The tip of the side-viewing needle transducer was placed into the tracheal tube, and the sample was rotated to produce an image of the trachea in the transverse plane. Following the experiment, images acquired from CPI-CTHI, PI-CTHI and F-CI are compared.

4. Results

From the simulation, PSFs were acquired and processed to generate two-dimensional images shown in Fig. 4 displayed in 60 dB dynamic range. In each column, three images acquired

by different methods – PI-CTHI, CPI-CTHI, F-CI – are sequentially displayed. Fig. 4a and b compares the images before and after applying the motor movement of 50 μm , the distance between two adjacent scan lines at their natural foci used in the measurement. Fig. 4a shows that range side lobes could not be identified without motor movement for all methods. With motor movement, Fig. 4b shows that artifacts mask PSF of PI-CTHI, while CPI-CTHI and F-CI present less range side lobes.

Fig. 5 shows axial and lateral profiles and spectral distribution of the PSFs shown in Fig. 4. Note that brightness profiles are acquired at 1.4 mm in axial and 0.65 mm in lateral direction. In this figure, blue⁴ dashed lines, solid red lines and black dot-dashed lines represent PI-CTHI, CPI-CTHI and F-CI respectively. Fig. 5a–c are simulated results without motor movement and Fig. 5d–f with motor movement. In the axial profiles, RSLL of PI-CTHI has increased from –58 dB to –37 dB with the movement while that of CPI-CTHI from –68 dB to –58 dB. Axial resolution value of all the methods is around 53 μm which is not affected by the transducer movement. The –6 dB beam width of PI-CTHI is the narrowest (178 μm) among the three PSFs when there is no motor movement while the others show the same width of 350 μm . Theoretically, the location of the side lobe, which can be defined from the center frequency and aperture size of a transducer [1], is between 150 μm and 300 μm laterally away from the beam center. Therefore, sidelobe reduction of PI-CTHI results in superior lateral resolution over either F-CI or CPI-CTHI. With motor movement, the –6 dB beam width is also maintained for all three methods. In the frequency domain, PI-CTHI has a 40 dB increase of power spectrum at 20 MHz, caused by fundamental leakages, under moving condition compared to that with no motor movement.

Fig. 6 shows variations of RSLL as a function of ‘*Weighting*’ values. With no motor movement, RSLL is maintained without being affected by the changes on ‘*Weighting*’. However, RSLL increases from –65 dB to –27 dB for CPI-CTHI as the ‘*Weighting*’ changes from –20 dB to 20 dB, and for PI-CTHI it is 21 dB higher when the motor is moved by 50 μm .

Fig. 7 shows the values of RSLL as the moving distance of a transducer is changed with 0 dB ‘*Weighting*’. The differences between the RSLL of PI-CTHI and CPI-CTHI are also plotted as triangular dots. After 10 μm travel distance, the RSLL rises to –49 dB and –35 dB for CPI-CTHI and PI-CTHI respectively because the correlation between adjacent scan lines with movement becomes weaker than with no-movement. The rate of increase, inferred from the increasing ‘*Difference*’ values, is smaller for CPI-CTHI than PI-CTHI.

Fig. 8 shows that RSLL changes depend on the value of non-linear coefficients, B/A , for CPI-CTHI (dashed line) and PI-CTHI (solid line). The values of RSLL for CPI-CTHI are not affected by the variation of the non-linear coefficient, while an increase of RSLL can be observed in PI-CTHI as the coefficient decreases with motor movement.

Fig. 9 shows difference values between the peak PSF amplitude of PI-CTHI and F-CI as a function of non-linear coefficients. Note that the transmitted pressure is 1 MPa for the both methods. The amplitude of harmonic signal keeps increasing up to a level 7 dB less than the fundamental signal when the B/A value becomes 10.2.

Fig. 10a–c shows experimentally acquired wire target images by CPI-CTHI, PI-CTHI and F-CI respectively displayed in 55 dB dynamic range. Range side lobes, which can be identified at positions above and below the main lobes in PI-CTHI image, are suppressed in the CPI-

⁴For interpretation of color in Figs. 2, 5 and 11, the reader is referred to the web version of this article.

CTHI image. The calculated eSNR is 60 dB for the CPI-CTHI, while for PI-CTHI and F-CI, they are 53 dB and 59 dB respectively.

Fig. 11a–c compares axial profiles and lateral profiles of the center wires and frequency responses of a scanline including the brightest spots for each imaging method. The –6 dB axial resolution of CPI-CTHI and PI-CTHI are the same at 75 μm . In the axial direction, RSL of PI-CTHI is –22 dB while CPI-CTHI –34 dB. The level of side lobes from the second wire is –30 dB for F-CI and reduced to –40 dB using CPI-CTHI. In the lateral direction, the beam size of CPI-CTHI is 110 μm which is same or slightly better than PI-CTHI (110 μm) and the F-CI (114 μm), while side lobe levels of CPI-CTHI (–18 dB) are between those of PI-CTHI (–40 dB) and F-CI (–15 dB). In the frequency domain, leakages of fundamental signals are identified. PI-CTHI has 10 dB higher level of leakages than CPI-CTHI at 25 MHz where the peak fundamental leakage is identified. The acquired eSNR of the CPI-CTHI from the wire phantom study is 60 dB. Compared with PI-CTHI, an improvement of 7 dB was achieved in eSNR. The measured results of the wire target studies are given in Table 2.

Fig. 12a–c shows the acquired cross sectional ultrasound images of a rabbit trachea by CPI-CTHI, PI-CTHI and F-CI respectively. Fig. 11d shows the histology of the trachea of an amphibian which has structures similar to Mammals' [18]. In the CPI-CTHI image, epithelial layers and connected tissues are clearly delineated, and blood vessels under the epithelial layers are better visualized than the images employing other methods. Particularly, a layer, a false echo caused by range side lobes, can be identified in PI-CTHI image in the outside of the cartilage layer, while not in the CPI-CTHI image and the blurry presentation inhibits the visualization of its internal structures. The contrast of the blood vessel, found in the upper left side of each ultrasound image, compared with brightness of the adjacent tissue is 6.0 dB for CPI-CTHI while 1.1 dB and 1.5 dB for PI-CTHI and F-CI respectively.

5. Discussion

The simulation data shown in Fig. 7 indicate that CPI-CTHI is capable of suppressing the RSL caused by the translation of a transducer. For a distance of 60 μm , which is close to the half of the beam width, between incident positions of a pulse inversion pair, the RSL for CPI-CTHI is still 10 dB less than the one for PI-CTHI. To make RSL in CPI-CTHI smaller than PI-CTHI and tissue contrast in CPI-CTHI better than F-CI, the rate of harmonic and fundamental signal should be optimized by adjusting '*Weighting*' value. The RSL suppression effect becomes optimized when –20 dB of '*Weighting*' is employed (–65 dB of RSL), sacrificing benefits of harmonic imaging, i.e., sidelobe suppression and tissue contrast enhancement. On the other hand, the increase of harmonic signals on CPI-CTHI elevates RSL up to –27 dB although the sidelobe level is the lowest among the three imaging approaches. With 0 dB '*Weighting*', RSL of –53 dB, which may be considered acceptable, higher than 45 dB, for medical ultrasound imaging [19] is achieved.

The '*Weighting*' values may be adjusted depending on imaging applications where the media have different non-linear coefficients. Previous research [20] show that B/A varies from 5 to 10 depending on the tissue types. In Fig. 9, the effectiveness of harmonic signals on CPI-CTHI is simulated by measuring the peak PSF amplitude of PI-CTHI compared with F-CI. The harmonic signals are increased at a level of 7 dB less than the fundamental signal with a B/A value of 10.2 although the contribution of harmonic signal is –20 dB compared to the fundamental signal with B/A value of 5.2. Therefore, harmonic signal of CPI-CTHI becomes stronger in a strong harmonic signal generating region like a fatty tissue where false echoes cause by range sidelobes would appear with PI-CTHI only mode, whereas

fundamental signals become dominant in the echo signals from a weak harmonic signal generator.

In the wire target study, the experimental results showed that the amount of RSLR suppression with CPI-CTHI compared to PI-CTHI is 12 dB while the simulation performed 21 dB. This discrepancy may be the result of the fact that the signal distortion caused by non-symmetric Gaussian spectral characteristics of the transducer employed and that non-uniform nonlinear coefficient distribution was not considered in the simulation. In addition to the RSLR reduction inherited from the F-CI, CPI-CTHI also possesses harmonic imaging characteristics in rejecting side lobes as shown in Fig. 9. Due to the effect of side lobe rejection, the -12 dB lateral resolution was improved from 270 μm to 220 μm compared with F-CI.

The rabbit trachea images clearly illustrate the advantages of CPI-CTHI. CPI-CTHI yields smaller RSLR near cartilages, higher echogenicity, and lower background noise level which make the blood vessel appear darker with an enhanced eSNR over either F-CI or PI-CTHI. The reason for this enhancement is that blood vessel is composed of fatty and muscular regions where harmonic effect of smaller beam width than F-CI may be expected.

6. Conclusion

Previously, pulse inversion chirp coded tissue harmonic imaging was typically not used for B-mode imaging due to the elevated range side lobe level. In the current study, combined pulse inversion chirp coded harmonic imaging is implemented to compensate for the shortcomings of both harmonic and fundamental imaging. The simulation study and experimental phantom study demonstrate that this new combined imaging approach is capable of achieving smaller range side lobe levels and better echo signal to noise ratio than harmonic only coded excitation imaging with a proto-typed intravascular ultrasound imaging system. The combined chirp coded tissue harmonic imaging also out performs fundamental chirp coded excitation imaging in reducing side lobes. The *ex vivo* study demonstrates that blood vessels of a rabbit trachea can be more clearly delineated by the combined imaging approach due to the non-uniform distribution of non-linear coefficients within the rabbit trachea.

Acknowledgments

This work has been supported by NIH Grants R01-HL79976 and P41-EB2182. The meanings of acronyms are provided on Table 3.

References

1. Shung, KK. Diagnostic Ultrasound: Imaging and Blood Flow Measurement. Taylor & Francis/CRC Press; Boca Raton: 2006. p. 148-149.
2. Hohl C, Schmidt T, Haage P, Honnef D, Blaum M, Staatz G, Guenther RW. Phase-inversion tissue harmonic imaging compared with conventional B-mode ultrasound in the evaluation of pancreatic lesions. *Eur. Radiol.* 2004; 14:1109–1117. [PubMed: 14714139]
3. Averkiou MA. Tissue harmonic imaging. *IEEE Ultrasonics Symposium.* 2000; 2:1563–1572.
4. Chiao RY, Hao X. Coded excitation for diagnostic ultrasound: a system developer's perspective. *IEEE Ultrason. Ferroelect. Freq. Contr.* 2005; 52:160–170.
5. Goertz DE, Frijlink ME, de Jong N, van der Steen AFW. Nonlinear intravascular ultrasound contrast imaging. *Ultrasound Med. Biol.* 2006; 32:491–502. [PubMed: 16616596]
6. Frijlink ME, Goertz DE, Vos HJ, Tesselaar E, Blacquiere G, Gisolf A, Krams R, van der Steen AFW. Harmonic intravascular ultrasound imaging with a dual frequency catheter. *Ultrasound Med. Biol.* 2006; 32:1649–1654. [PubMed: 17112951]

7. Frijlink ME, Goertz DE, de Jong N, van der Steen AFW. Pulse inversion sequences for mechanically scanned transducers. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.* 2008; 55:2154–2163.
8. Shen CC, Li PC. Motion artifacts of pulse inversion-based tissue harmonic imaging. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.* 2002; 49:1203–1211.
9. Bosisio MR, Hasquenoph JM, Sandrin L, Laugier P, Bridal SL, Yon S. Real-time chirp-coded imaging with a programmable ultrasound biomicroscope. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.* 2010; 57:654–664.
10. Kremkau FW, Taylor KJ. Artifacts in ultrasound imaging. *J. Ultrason. Med.* 1986; 5:227–237.
11. Feng N, Sun M, Ma L, Ma J. Curvelet based image fusion for ultrasound contrast harmonic imaging. *Int Conf Intelligent Info Hiding and Multimedia Signal Processing.* 2009:953–956.
12. Liu G, Shen Y, Wang Y, Wang Q. On the use of data fusion in ultrasonic imaging. *Instrumentation and Measurement Technology Conference.* 2004; 1:760–762.
13. Nowicki A, Wojcik J, Secomski W. Harmonic imaging using multitone nonlinear coding. *Ultrasound Med. Biol.* 2007; 33:1112–1122. [PubMed: 17451870]
14. Kanagaratnam P, Gogineni SP, Ramasami V, Braaten D. A wideband radar for high-resolution mapping of near-surface internal layers in glacial ice. *IEEE Trans. Geosci. Remote Sens.* 2005; 42:483–490.
15. Kim DY, Lee JC, Kwon SJ, Song TK. Ultrasound second harmonic imaging with a weighted chirp signal. *IEEE Ultrasonics Symposium.* 2001; 2:1477–1480.
16. Anderson ME. A 2D nonlinear wave propagation solver written in open-source matlab code. *IEEE Ultrasonics Symposium.* 2000; 2:1351–1354.
17. Madsen EL, Frank GR, McCormick MM, Deaner ME, Stiles TA. Anechoic sphere phantoms for estimating 3-d resolution of very-high-frequency ultrasound scanners. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.* 2010; 57:2284–2292.
18. Kuehne B, Junqueira LCU. Histology of the trachea and lung of *Siphonops annulatus* (Amphibia, Gymnophiona). *Rev. Bras. Biol.* 2000; 60:167–172. [PubMed: 10838935]
19. Haider B, Lewin PA, Thomenius KE. Pulse elongation and deconvolution filtering for medical ultrasonic imaging. *IEEE Trans. Ultrason. Ferroelect. Freq. Contr.* 1998; 45:98–113.
20. Sehgal CM, Brown GM, Bahn RC, Greenleaf JF. Measurement and use of acoustic nonlinearity and sound speed to estimate composition of excised livers. *Ultrasound Med. Biol.* 1986; 12:865–874. [PubMed: 3810981]

Fig. 1. Generation of a combined chirped waveform. Both harmonic and fundamental chirp are summed to generate the final waveform.

Fig. 2.
Block diagram of a custom-designed rotational mechanical scanner with a side-viewing needle transducer.

Fig. 3.
Block diagram of post processing used in UBM software.

Fig. 4. PSF from simulation studies (a) with no motor movement and (b) with motor movement. From the top image, PI-CTHI, CPI-CTHI and F-CI are displayed sequentially.

Fig. 5. Brightness profiles of PSFs with no motor movement along (a) axial (b) lateral direction and (c) spectra for PI-CTHI (blue dashed line), CPI-CTHI (red solid line), F-CI (black dash-dotted line). Brightness profiles of PSFs with motor movement along (d) axial (e) lateral direction and (f) spectra for PI-CTHI (blue dashed line), CPI-CTHI (red solid line), F-CI (black dash-dotted line).

Fig. 6. Changes of RSL depending on the '*Weighting*' values in the Eq. (3). '◆' and '■' indicate RSL for CPI-CTHI and for PI-CTHI respectively with no motor movement. '▲' and 'X' indicate RSL for CPI-CTHI and for PI-CTHI respectively with 50 μm motor movement.

Fig. 7.

Changes of RSLL depending on the travel distance with motor movement. '◆' and '■' indicate RSLL for CPI-CTHI and for PI-CTHI respectively. '▲' is differences of the RSLL between CPI-CTHI and PI-CTHI referring to the 'Difference (dB)' axis.

Fig. 8.

Changes of RSLL depend on the non-linear coefficient B/A value. '■', '◆' and '●' indicate the RSLL with the non-linear coefficient values of 2.2, 5.2 and 8.2 respectively. CPI-CTHI and PI-CTHI are plotted with dashed lines and solid lines respectively.

Fig. 9.
Changes of the differences between peak PSF amplitude values of PI-CTHI and F-CI depending of non-linear coefficients, B/A .

Fig. 10.

Wire target images acquired by (a) CPI-CTHI, (b) PI-CTHI and (c) FI. In the images, white arrows are indicating wire targets.

Fig. 11. Brightness profiles of the first wires along (a) axial direction and (b) lateral direction. (c) Frequency responses of a scanline having the brightest spot along the axial direction. Black dot-dashed line indicates FI, red solid line CPI-CTHI and blue dashed line PI-CTHI.

Fig. 12. Rabbit trachea ultrasound images acquired by (a) CPI-CTHI, (b) PI-CTHI and (c) FI. (d) Histology of trachea from an amphibian.

Table 1

Non linear simulation parameters.

Parameters	Values	Parameters	Values
Sound speed (c)	1500 m/s	Bandwidth of transducer	75%
Density of medium (ρ)	100 kg/m ³	F-number	2.9
Nonlinear coefficient (B/A)	2.2, 5.2, 8.2	Attenuation coefficient	0.5 dB/cm/MHz
Bulk modulus (K)	2.37 GPa	Chirp window type	Hanning
Peak pressure (p)	1 MPa	F-CI frequency	20–60 MHz
Center frequency	35 MHz	PI-CTHI frequency	10–30 MHz

Table 2

Quantitative results from phantom studies.

	CPI-CTHI	PI-CTHI	FI
–6 dB axial resolution	75 μm	75 μm	76 μm
–6 dB lateral resolution	149 μm	144 μm	200 μm
eSNR	60.0 dB	53 dB	59 dB
Sidelobe level	–18 dB	–25 dB	–15 dB
RSLL	–34 dB	–22 dB	N/A

Table 3

Acronyms.

Acronyms	Full name	Acronyms	Full name
RSLL	Range side lobe level	PSF	Point spread function
eSNR	Echo signal to noise ratio	PI-CTHI	Pulse inversion chirp coded tissue harmonic imaging
IVUS	Intravascular ultrasound	CPI-CTHI	Combined PI-CTHI
THI	Tissue harmonic imaging	F-CI	Fundamental chirp coded imaging
DI	De-ionized		