Article:

Arif, M, Ali, MA, Shaikh, MM et al. (2017) Investigation of Non-linear Chirp Coding for Improved Second Harmonic Pulse Compression. Ultrasound in Medicine & Biology, 43 (8). pp. 1690-1702. ISSN: 0301-5629

https://doi.org/10.1016/j.ultrasmedbio.2017.03.005

Investigation of Nonlinear Chirp Coding for Improved Second Harmonic Pulse Compression

Muhammad Arif[a,*], Muhammad Asim Ali[b], Muhammad Mujtaba Shaikh[c], Steven Freear[d], Corresponding Author[*]

[a] *Department of Biomedical Engineering, Mehran University of Engineering and Technology, Jamshoro, Pakistan*
[b] *Department of Electrical Engineering, Sukkur Institute of Business Adminstration, Sukkur, Pakistan*
[c] *Department of Basic Sciences and Related Studies, Mehran University of Engineering and Technology, Jamshoro, Pakistan*
[d] *School of Electronic and Electrical Engineering, University of Leeds, Leeds, UK*

Abstract

In this paper, a nonlinear frequency-modulated (NLFM) chirp coding was investigated to improve the pulse compression of the second harmonic chirp signal by reducing the range sidelobes level. The problem of spectral overlapping between fundamental and second harmonic component (SHC) was also investigated. Therefore, two methods were proposed, method-I show the scenario of non-overlap condition and method-II with pulse inversion technique was used for overlap harmonic condition. In both methods, the performance of the NLFM chirp was compared with the reference linear frequency-modulated (LFM) chirp signals. Experiments were performed using a 2.25

---

[*] Corresponding Author: Muhammad Arif
Postal Address: Department of Biomedical Engineering, Mehran University of Engineering & Technology, Jamshoro, 76062, Sindh, Pakistan
Phone: +92 (22) 2772250-73, Ext: 7016
Email: m.arif@faculty.muet.edu.pk
URL: https://sites.google.com/site/mdotarif/

MHz transducer mounted coaxially at a distance of 5 cm with a 1 mm hydrophone in a water tank and the peak negative pressure of 300 kPa was set at the receiver. Both simulations and experimental results show that the peak sidelobe level (PSL) of the compressed SHC of NFLM chirp was improved by at least 13 dB in method-I and 5 dB in method-II when compared with the PSL of LFM chirps. Similarly, the integrated sidelobe level (ISL) of the compressed SHC of NLFM chirp was improved by at least 8 dB when compared with the ISL of LFM chirps. In both methods, the axial mainlobe width of the compressed NLFM chirp was comparable to the LFM signals. The signal-to-noise ratio of the SHC of NLFM was improved by up to 0.8 dB, when compared with the SHC of LFM signal having the same energy level. Results were also presented which show the robustness of NLFM chirp under a frequency dependent attenuation of 0.5 dB/[cm×MHz] up to penetration depth of 5 cm and a Doppler shift of up to 12 kHz.

Keywords: Ultrasound, Nonlinear chirp, Pulse compression, Pulse inversion, Harmonic imaging

Introduction

In recent years, ultrasound harmonic imaging has become prevalent in commercial medical ultrasound imaging systems. Ultrasound harmonic imaging relies on the second or higher order harmonic components. In tissue harmonic imaging, these nonlinear harmonics are produced by finite amplitude distortion of ultrasound waves propagating through biological tissue (Duck, 2010). Whereas in ultrasound contrast imaging, these harmonics are produced by the nonlinear scattering from contrast microbubbles (de Jong et al., 2002; Maresca et al., 2014). Ultrasound images based on the nonlinear second harmonic component (SHC) provide improved spatial resolution with reduced reverberation artifacts when compared to conventional (fundamental) B-mode imaging (Jensen, 2007; Wells, 2006).

Coded excitation techniques were originally introduced in radar communication and now are widely used in medical ultrasound imaging systems to provide improved signal-to-noise ratio (SNR) (Cook and Bernfeld, 1967; Chiao and Hao, 2005). Coded excitation with long duration linear frequency modulated (LFM) chirp signals offer the potential to improve the SNR of the SHC. This can be done without increasing the peak excitation pressure or mechanical index (MI) and without reducing the system frame-rate (Cobbold, 2007). However, on the receiving side of the system, harmonic matched filters are typically used to extract and compress the SHC and to recover signal axial resolution (Kim et al., 2001; Arif et al., 2010a). The SNR of a chirp signal depends on the time-bandwidth product (TBP) and can also be improved by extending the signal bandwidth. However, in ultrasound harmonic imaging application, the signal bandwidth extension is restricted by the finite bandwidth of the ultrasound transducer to accommodate both the fundamental and the SHC, and the spectral overlapping between the fundamental and the SHC (Averkiou, 2000).

The power spectrum of an unweighted LFM chirp is approximately rectangular in shape and yields a sinc-like function after pulse compression. The compressed chirp signal contains a peak sidelobe level (PSL) at ∼ -13 dB. This higher value of PSL will mask out the mainlobe width (MLW) from the weak scatterer and will potentially degrade the image contrast by appearing as false echoes. Therefore, the higher values of the PSL are unacceptable in modern medical ultrasound imaging systems operating at a dynamic range of more than 60 dB (Johnston and Fairhead, 1986; Misaridis and Jensen, 2005b).

In order to reduce the higher PSL of the compressed chirp signal, a strong weighting function is applied either on the transmitting signal or on the received matched filter; the latter case is termed a mismatched filter. Windowing on the excitation signal causes a reduction in the transmitting energy and hence penetration depth, whilst windowing on the matched filter results in reduced gain in the SNR and axial resolution. Therefore, a tradeoff between the MLW and PSL is exist in the pulse compression process of the LFM signal (Adams, 1991; Milleit, 1970).

Nonlinear frequency modulated (NLFM) chirp signals provide an alternative means to modify the rectangular power spectrum of the LFM chirp into a desirable shape. The NLFM chirp can be designed to optimise the signal transmitting energy and the shape of the power spectrum so that it matches spectrally with the transfer function of the transducer. This results in more energy transmitted though the transducer which potentially improve the SNR and penetration depth. Also, reduced PSL will be get after pulse compression without using any additional windowing function on the matched filter (Harput et al., 2013; Arif et al., 2010b; Gran and Jensen, 2007; Pollakowski and Ermert, 1994).

The effects of shaping the transmitting spectrum using the NLFM chirp for improved spectral matching with the transmitter were first studied by PS. Brandon (Brandon, 1973). He had designed the nonlinear pulse compression system for radar to get the high resolution with reduced loss in the SNR. The NLFM chirp was designed using the least squares optimization method for synthetic transmit aperture B-mode imaging (Gran and Jensen, 2007). The NLFM signal with the quadratic instantaneous frequency function was designed and implemented for tissue harmonic imaging (Song et al., 2011).

In this paper, NLFM chirp coding was investigated as an excitation scheme in the area of ultrasound harmonic imaging. The aim was to reduce the PSL after pulse compression and to improve the SNR of the second harmonic chirp component.

The remaining sections of the paper are divided as follows: Section-II describes the basic theory and design methods of NLFM and reference LFM chirp signals, Section-III describes about the proposed methods, Section-IV describes the simulation and experimental procedures with the post processing of nonlinear received signals, the results of harmonic pulse compression and Doppler sensitivity evaluation of designed chirp signals are examined in Section-V, and finally the performance of second harmonic pulse compression achieved using the Nonlinear chirp coding is discussed in Section-VI.

Theory and Signal Design

Nonlinear Frequency Modulated (NLFM) Signals

In exponential form, the time domain chirp signal $x\left(t\right)$ can be expressed as Misaridis and Jensen (2005a),

$$x\left(t\right) = p\left(t\right)e^{j2\pi\phi\left(t\right)} \tag{1}$$

where $p\left(t\right)$ and $\phi\left(t\right)$ are the amplitude and phase modulation functions of the chirp signal, respectively.

The spectrum of the chirp signal given in (1) is expressed as,

$$X(\omega) = \int_{-\infty}^{\infty} p\left(t\right)e^{j\{-\omega t+\phi(t)\}}dt \tag{2}$$

The integrand $[-\omega t+\phi(t)]$ in (2) is an oscillating function which is varying at the rate of $\frac{d}{dt}[-\omega t+\phi(t)]$. The major contribution to the chirp spectrum occurs when the rate of change of oscillating function is minimal and is also referred to as the stationary phase point, this can be expressed as

$$\frac{d}{dt}[-\omega t+\phi(t)] = 0 \tag{3}$$

In (3), ω and t are two independent variables. Therefore, the particular value of t that can satisfy the condition given in (3) can be found by assuming the value of ω. In the case of an LFM signal with a quadratic phase modulation function, the integral in (2) can be solved analytically. However, in the case of nonlinear phase modulation function, the second order Taylor expansion of phase $\phi(t)$ around t_k which is the solution of the (3) at ω_k is used to reduce the integral in (2) to Fresnel integral. Hence after algebraic manipulation, the power spectrum of the chirp signal at ω_k is given by (Cook and Bernfeld, 1967),(Collins and Atkins, 1999), (Arif, 2010)

$$|X(\omega_k)|^2 \approx 2\pi \frac{p^2(t_k)}{|a(t_k)|} \tag{4}$$

where $|X(\omega_k)|^2$ is the power spectrum of the chirp signal, $p(t_k)$ is the amplitude modulation function, and $a(t_k)$ is the chirp-rate function of the signal.

In the case of an LFM signal, the term chirp-rate function $a(t_k)$ in (4) is constant and the shape of the power spectrum is controlled by modifying the amplitude modulation function $p(t_k)$ of the signal using a time or frequency domain windowing function. However, in the case of the NLFM signal, the shape of the power spectrum is controlled by either modifying the chirp-rate function and keeping the amplitude modulation function constant (rectangular envelope) or by modifying both the chirp-rate and amplitude modulation functions of the signal; the latter case is known as a hybrid design approach. The main advantage of the hybrid design approach is that the designed NLFM signal will be less sensitive to a Doppler shift caused by the moving blood or tissue. However, this advantage comes at the expense of reduced SNR of the transmitting signal due to amplitude tapering (Johnston and Fairhead, 1986).

In a chirp coded excitation system, a matched filter is used on the receiving side to perform pulse compression of the long duration signal to recover axial resolution. The matched filter will cancel out the phase information of the chirp signal and therefore phase has no contribution in the axial resolution. The output of the matched filter is the autocorrelation function (ACF) of the chirp signal (Girod et al., 2001).

$$R_{xx}(\tau) = \int_{-\infty}^{\infty} x(t+\tau)x^*(t)dt \tag{5}$$

Also the ACF of the chirp signal is the inverse Fourier transform of the power spectrum and is expressed as

$$\begin{aligned} R_{xx}(\tau) &= \int_{-\infty}^{\infty} \left\{ |X(f)| e^{j\phi(f)} \right\} \left\{ |X(f)| e^{-j\phi(f)} \right\} e^{j2\pi f\tau} df \\ &= \int_{-\infty}^{\infty} |X(f)|^2 e^{j2\pi f\tau} df \end{aligned} \tag{6}$$

Therefore, the mainlobe width and sidelobe level of the compressed chirp signal will depend on the shape of the power spectrum. Hence choosing a suitable window function as a shape of the power spectrum would yield a low sidelobe level after pulse compression. In this paper, NLFM signals were designed using the hybrid approach and a tapered cosine window (also known as a Tukey window) was selected as a shape of the desired power spectrum and is expressed as (Harris, 1978),

$$|X(\omega)|^2 = \begin{cases} 1, & 0 \leq |\omega| \leq \beta\frac{B}{2} \\ \frac{1}{2}\left[1 + \cos\left(\pi\frac{\omega - \beta\frac{B}{2}}{2(1-\beta)\frac{B}{2}}\right)\right], & \beta\frac{B}{2} \leq |\omega| \leq \frac{B}{2} \end{cases} \tag{7}$$

where B is the bandwidth of $X(\omega)$, and β is the envelope tapering ratio: $\{0 \leq \beta \leq 1\}$.

The tapered cosine window with a tapering ratio of 100% ($\beta = 1$) was carefully chosen for two reasons. Firstly, it matches with the transfer function of the ultrasound transducer used in the experiments and secondly it provides reduced PSL after pulse compression of the SHC of the received NLFM signal.

The nonlinear instantaneous frequency function, $f_i\left(t\right)$, of the NLFM signal is expressed as (Collins and Atkins, 1999),

$$f_i\left(t\right) = f_c + \frac{B}{2}\left[\frac{\alpha\tan\left(\frac{2\gamma t}{T}\right)}{\tan(\gamma)} + \frac{2\left(1-\alpha\right)t}{T}\right] \tag{8}$$

where the parameters α and γ are used to control the nonlinear curve of the instantaneous frequency function. B is the sweeping bandwidth, f_c is the centre frequency, and T is the chirp duration.

The chirp-rate function, $a(t)$, of the NLFM signal was found by taking the derivative of the instantaneous frequency function, $f_i\left(t\right)$, in (8).

$$a(t) = \frac{d}{dt}\left(f_i\left(t\right)\right) = \frac{B}{T}\left[\frac{\alpha\gamma\left(1+\tan^2\left(\frac{2\gamma t}{T}\right)\right)}{\tan(\gamma)} + \left(1-\alpha\right)\right] \tag{9}$$

The amplitude modulation function, $p(t)$, of the NLFM signal was obtained by rearranging (4) and substituting the values of power spectrum and chirp-rate function from (7) and (9).

$$p\left(t_k\right) \approx \sqrt{\left|X(\omega_k)\right|^2\left|a\left(t_k\right)\right|} \tag{10}$$

The phase modulation function, $\phi\left(t\right)$, of the NLFM signal was obtained by taking the integral of the instantaneous frequency function, $f_i\left(t\right)$, in (8) with respect to time t.

$$\phi\left(t\right) = \int_0^t f_i\left(t\right)dt,\ 0 \leq t \leq T \tag{11}$$

Finally, the NLFM signal was found by substituting the values of the amplitude modulation function $p(t)$, and the phase modulation function $\phi(t)$ from (10) and (11) into (1).

Linear Frequency Modulated (LFM) Signals

The LFM signals were designed using equation (1), where the phase modulation function of the LFM signal is expressed as:

$$\phi\left(t\right)=\frac{B}{2T}t^2+(f_c-\frac{B}{2})t+\varphi,\ 0\leq t\leq T \tag{12}$$

where B is the sweeping bandwidth, T is the chirp duration, f_c is the centre frequency, and φ is the initial phase of the chirp signal and have values of 0 and 0.5 for pulse inversion.

In this study, two LFM signals were used as a reference excitation. Each signal had a duration (T) of 20 μs, and a centre frequency (f_c) of 2.25 MHz. The difference in each LFM signal was the transmitting energy which was set by the application of tapered cosine window with envelope tapering ratios (β) of 10% and 80%. Respectively after the application of a windowing function the LFM signals will be termed as LFM-W10 and LFM-W80. The energy of the LFM-W80 signal was set equal to the NLFM signal whereas the energy of the LFM-W10 signal was 43% higher than the NLFM signal. The LFM-W10 and LFM-W80 signals were designed to show the effect of different windowing on the transmitting energy, power spectrum and harmonic pulse compression.

Excitation Signals

The design parameters of the NLFM and reference LFM excitation signals are shown in Table 1. The instantaneous frequency functions, $f_i\left(t\right)$, of the NLFM, LFM-W80, and LFM-W10 signals are shown in Fig. 1(Top). The instantaneous frequency function of the NLFM signal is nonlinear and is similar in appearance to a reverse ‘S’ shape. However, the instantaneous frequency functions of the LFM signals are linear. The NLFM and LFM signals have equal sweeping bandwidth (B) and time duration (T).

The chirp-rate function, $a(t)$, of the NLFM signal shows higher values at the edges and lower values in the middle part of the signal and appears as a ‘U’ shape. However, the chirp-rate functions of the LFM signals are constant throughout the signal’s duration as shown in Fig. 1(Bottom). Note that both the instantaneous frequency and chirp-rate functions of LFM signals have equal magnitude because of the same sweeping bandwidth and duration.

The amplitude envelopes, $p(t)$, of the NLFM, LFM-W80, and LFM-W10 excitation signals are shown in Fig. 2(Top). The amplitude envelope of the LFM-W80 chirp was designed so that it can transmit acoustical energy equal to the NLFM signal. The amplitude envelope of the LFM-W10 chirp is nearly flat throughout the signal duration. This is due to the fact that less amplitude tapering ($\beta = 0.1$) was applied. Therefore, the LFM-W10 signal can transmit 43% more acoustical energy than the NLFM and LFM-W80 signals.

The power spectra of the NLFM, LFM-W80, and LFM-W10 signals are shown in Fig. 2(Bottom). The transfer function (TF) of the 2.25 MHz transducer used in the experiment is also shown in the figure. The TF of the transducer was accounted in the design process of the NLFM signal. The NLFM and reference LFM signals have same sweeping bandwidth. However, their -6 dB bandwidths were different due to the different amplitude tapering used in the design process. The power spectrum of the LFM-W10 signal contains Fresnel-ripples in the pass band due to less amplitude tapering ($\beta =$

0.1) and their -6 dB bandwidth was 40% higher than the NLFM signal. Similarly, the power spectrum of the LFM-W80 signal is smooth and contains no ripples in the pass band. However, their -6 dB bandwidth was 11% higher than the NLFM signal.

Proposed Methods

In this paper, two methods were proposed to show the scenarios of non-overlap and overlap harmonic conditions. The proposed system flow charts are shown in Fig. 3.

Method I: For Non-Overlap Harmonic Condition

Method I require a single transmission of a chirp signal with a fractional bandwidth (FBW) of 20%. In the power spectrum of the received signal, the nonlinear spectral harmonics do not overlap with each other due to the use of narrow bandwidth chirp excitation. Therefore, the application of a harmonic matched filter would yield the extraction and compression of the second harmonic chirp component. Note that the bandpass filter response is inherent within the harmonic matched filter specification as shown in Fig. 3(Top).

Method II: For Overlap Harmonic Condition

In ultrasound harmonic imaging, the bandwidth of the SHC is twice the bandwidth of the fundamental frequency component Chiao and Hao (2005); Misaridis and Jensen (2005b). Therefore, increasing the bandwidth of an excitation signal will cause spectral overlapping between the fundamental and the SHC. The direct application of a second harmonic matched filter under an overlap harmonic condition will result in the production of higher range sidelobes after pulse compression. Therefore, in this method before pulse compression, the pulse inversion (PI) technique was proposed to extract the SHC under an overlap harmonic condition. PI requires two transmissions of 45% fractional bandwidth chirp signals with a phase difference of 180°. The summation of the corresponding received echoes will cancel out the fundamental component and will extract the SHC under an overlap harmonic condition (Simpson et al., 1999; Ma et al., 2005). After the extraction of the SHC, the application of a harmonic matched filter would yield the pulse compression of the second harmonic chirp component as shown in Fig. 3(Bottom).

Simulation and Experimental Investigation

Simulations

The nonlinear propagation of ultrasound waves in an attenuating medium was simulated using the following nonlinear model (Wojcik et al., 1998).

$$\rho\frac{\partial^2 u}{\partial t^2} = -\nabla p,\ p = -\rho c^2\left(\nabla \cdot u + \frac{1}{2}\frac{B}{A}(\nabla \cdot u)^2\right) \tag{13}$$

where u is the particle displacement vector, c is the speed of sound, p is the acoustic pressure, ρ is the material density, and B/A is the nonlinear coefficient.

The model was simulated in MATLAB (The MathWorks Inc., Natick, MA, USA) using the time-domain based pseudo-spectral method. A detailed description of the simulation method can be found in (Anderson, 2000). The simulation model was noise free and does not account for any transducer geometry, therefore only plane waves were applied in simulations.

The performance of the pulse compression system is also affected by the spectral mismatched of the receiving filter. This is caused by the frequency dependent attenuation of tissue, which also limit its implementation in real time imaging systems (Ramalli et al., 2015). In the case of ultrasonic attenuation, the power spectrum of the backscattered signal is frequency shifted and distorted as a function of signal bandwidth and propagation depth. The higher frequencies in the bandwidth are more attenuated than the lower frequencies. This results in a reduction of the effective TBP of the signal and hence the SNR (Rao, 1994). Therefore, the effect of frequency dependent attenuation as a function of bandwidth was also investigated in all simulations. The frequency dependent attenuation of 0.5 dB/[cm×MHz] was considered up to penetration depth of 5 cm. The simulation parameters are shown in Table 2, (Hallaj et al., 2001).

Experimental Setup and Procedure

In order to validate the proposed methods, experiments were performed in a tank containing deionised, degassed filtered water at the temperature of 21 ± 1°C. The schematic diagram of the experimental setup is shown in Fig. 4. The NLFM and reference LFM excitation signals were designed in Matlab (The MathWorks Inc., Natick, MA, USA) and then loaded into a programmable arbitrary waveform generator (AWG) (33250A Agilent Technologies Inc., 80 MHz, Santa Clara, CA, USA). The generated signals from AWG were then amplified using an RF power amplifier (A150 Electronics & Innovation Ltd., 55 dB, Rochester, NY, USA). The amplified signals were then applied to a single element immersion transducer (V323-SM, Olympus NDT Inc., Waltham, MA, USA) with a pulse repetition frequency of 100 Hz. The transducer has an active element diameter of 6.35 mm, central frequency of 2.25 MHz with a -6 dB FBW of 56%, and a focal length is in between 8.9 mm to 11.4 mm. The transducer was mounted coaxially with a hydrophone at a distance of 50 ± 0.1mm. All measurements were performed in the far field of the transducer. The distance and alignment between the transducer and hydrophone were controlled using a custom built computer numerical control (CNC) scan system. A needle-type Polyvinylidene Fluoride (PVDF) hydrophone with an active element diameter of 1.0 mm (calibrated from 1 to 20 MHz, Precision Acoustics Ltd., Dorchester, UK) was used to receive the nonlinear signals. The pressure level of each waveform was calibrated and a mechanical index (MI) of 0.2 (peak negative pressure of 300 kPa at 2.25 MHz) was received by the hydrophone. The average noise measurement for the overall system was performed before starting the experiments. This was done with the same experimental setup except that the AWG output being switched off. The signals received from the hydrophone were captured 64 times and averaged using a digital oscilloscope (44Xi LeCroy Corporation, 400 MHz, Chestnut Ridge, NY, USA) at a sampling frequency of 100 MS/s with 8-bit resolution. All the acquired data were stored in a computer system and processed off-line using MATLAB (The MathWorks Inc., Natick, MA, USA). All the received signals were corrected using an inverse filter designed according to the frequency response of the hydrophone where the calibration data were supplied by the manufacturer (Precision Acoustics Ltd., Dorchester, UK).

Post Processing of Nonlinear Received Signals

NLFM Received Signals

In the case of the NLFM signal, the shaping of the power spectrum was done on the transmitting side. Therefore on the receiving side of the system, the harmonic matched filter was used to perform pulse compression of the SHC of the received signal. The impulse response of the harmonic matched filter was the time-reversed conjugate of the NLFM chirp with twice the centre frequency and bandwidth parameters so that it matches with the SHC of the received signal. Also the same windowing function was used in the design of the harmonic matched filter as used in the excitation signal (Borsboom et al., 2005).

LFM Received Signals

Since a strong amplitude tapering ($\beta = 0.8$) was used in the design process of the LFM-W80 signal, therefore a harmonic matched filter was used to perform pulse compression of the SHC of the received signal. However, in the case of the LFM-W10 signal, less amplitude tapering ($\beta = 0.1$) was applied to the excitation signal. Hence, instead of a harmonic matched filter, a harmonic mismatched filter was used on the receiving side to perform second harmonic pulse compression. The harmonic mismatched filter was designed by applying a Chebyshev window of 100 dB attenuation to the harmonic matched filter. This results in the further reduction of PSL after pulse compression.

Performance Evaluation of Second Harmonic Pulse Compression

The performance of second harmonic pulse compression of NLFM, LFM-W80 and LFM-W10 received signals was quantitatively assessed by using the three quality parameters, namely: the mainlobe width (MLW), peak sidelobe level (PSL), and integrated sidelobe level (ISL). In the compressed second harmonic chirp signal, the axial resolution was evaluated by measuring the MLW at -20 dB in microseconds. The PSL and ISL provide a measure of self induced noise and contrast resolution. The PSL was evaluated by taking the ratio of the highest sidelobes peak to mainlobe peak in decibels. The ISL was computed by taking the ratio of total sidelobes power to the mainlobe power in decibels (Arshadi et al., 2007), (Misaridis et al., 2000).

Results and Discussion

Simulation Results: Method I

The power spectra of the simulated nonlinear received signals with a FBW of 20% are shown in Fig. 5. The SHC of the received NLFM, LFM-W80 and LFM-W10 signals were respectively -23.5 dB, -23.8 dB and -23.3 dB below the fundamental frequency component. The SHC of NLFM was improved by 0.3 dB when compared with the SHC of the LFM-W80 signal of same energy level. The SHC of NLFM was 0.2 dB below with the SHC of the LFM-W10 signal.

The envelopes of the simulated compressed second harmonic chirp signals are shown in Fig. 6. The pulse compression results of second harmonic chirp signals are shown in Table 3. The results showed that at least 15 dB improvement in the PSL of the compressed second harmonic signal of the NLFM when compared with the PSL of LFM-W80 and LFM-W10 signals. The ISL of the compressed NLFM signal was improved by 9 dB when compared with the ISL of the compressed LFM-W80 signal. Similarly the ISL of the compressed NLFM signal was improved by 13.9 dB when compared with the ISL of the compressed LFM-W10 signal. The MLW of the compressed second harmonic signal of NLFM was comparable to the MLW of LFM-W80 and LFM-W10 signals.

Simulation Results: Method II

The power spectra of the simulated nonlinear received NLFM, LFM-W80 and LFM-W10 signals with a FBW of 45% are shown in Fig. 7. The SHC of the NLFM was -24.1 dB, whereas the SHC of the LFM-W80 and LFM-10 received signals were respectively -25.0 dB and -23.5 dB below the fundamental frequency component. The SHC of NLFM was improved by 0.9 dB when compared with the SHC of the LFM-W80 signal of same energy level. The SHC of NLFM was 0.6 dB below with the SHC of the LFM-W10 signal. The figure shows the spectral overlapping of the second harmonic with the fundamental and third harmonic components. The PI technique was applied to these nonlinear received signals in order to extract the SHC. After PI processing the SHC of NLFM, LFM-W80 and LFM-W10 signals were improved by 12 dB.

The envelopes of the simulated compressed second harmonic chirp signals after processed with the PI are shown in Fig. 8. The results indicated that the PSL of the compressed second harmonic signal of the NLFM was improved by 27.6 dB and 15 dB when compared with the PSL of LFM-W80 and LFM-W10 signals, respectively. Similarly the ISL of the compressed second harmonic signal of the NLFM was improved by 13.6 dB and 12.8 dB when compared with the ISL of LFM-W80 and LFM-W10 signals, respectively. The MLW of the NLFM compressed second harmonic signal was higher, but comparable to the LFM-W80 and LFM-10 signals.

Experimental Results: Method I

The power spectra of the measured nonlinear received signals at a depth of 5 cm are shown in Fig. 9. The NLFM and LFM spectrum plots were compared by normalising each spectrum with its own maximum value so that the fundamental component of each signal was aligned to the 0 dB. The figure shows the existence of the fundamental and nonlinear second and third harmonic components. These nonlinear harmonic components were produced due to the nonlinear propagation of ultrasound waves through water at the higher acoustic pressure exerted by the transducer. The spectra of the received signals show that the nonlinear harmonic components do not overlap with each other due to the use of 20% FBW excitation. The third harmonic component exists in the received signal because a broadband hydrophone was used as a receiver. In a clinical setting for an ultrasound second harmonic imaging, the third harmonic component will be filtered out or further suppressed by a limited bandwidth of the receiving transducer.

In the frequency domain, the SHC of the received NLFM, LFM-W80 and LFM-W10 signals were respectively -18.8 dB, -19.6 dB and -18.4 dB below the fundamental frequency component. The SNR was estimated in the frequency domain by measuring the peak height of the second harmonic relative to the mean noise value. The SNR of the SHC of NLFM was improved by 0.8 dB when compared with the SHC of the LFM-W80 signal. The 0.8 dB improvement in the SNR was small because both signals contained the same energy level, sweeping bandwidth, and duration. The SNR of the SHC of NLFM was comparable with the SHC of the LFM-W10 signal even though the NLFM signal contained 57% of the energy of the LFM-W10 signal.

The envelopes of the compressed second harmonic chirp signals are shown in Fig. 10. The pulse compression results of the SHC are shown in Table 4. The PSL of the compressed second harmonic signal of the NLFM was improved by 13 dB when compared with the PSL of the LFM-W80 signal. Similarly the PSL of the compressed second harmonic signal of the NLFM was improved by 15.6 dB when compared with the PSL of the LFM-W10 signal. The compressed second harmonic signal of the LFM-W10 contains higher PSL due to the existence of Fresnel ripples in the power spectrum. Another reason for the higher PSL is the slight overlapping of the SHC with the fundamental and third harmonics. This occurs as the LFM-W10 signal contains 40% higher -6 dB bandwidth than the NLFM signal due to less amplitude tapering. The ISL of the compressed second harmonic signal of the NLFM was improved by 8.4 dB and 18.6 dB when compared with the ISL of LFM-W80 and LFM-W10 signals, respectively. The improvement in PSL and ISL will potentially improve the image quality and increase the contrast resolution.

By comparing the MLW of the compressed second harmonic chirp signals it was found that the compressed NLFM chirp had a slightly higher value of MLW. This result was expected due to the fact that the NLFM chirp had 11% and 40% lower -6 dB bandwidth than the LFM-W80 and LFM-W10 signals, respectively. The MLW of the compressed chirp signal is inversely proportional to the bandwidth of the signal. Therefore, in a pulse-echo mode for ultrasound harmonic imaging, the MLW of the compressed second harmonic NLFM and LFM chirps will be further increased due to the bandwidth reduction of the SHC caused by the frequency dependent attenuation in tissue and limited bandwidth of the receiving transducer.

Experimental Results: Method II

The power spectra of the measured nonlinear received signals with a FBW of 45% are shown in Fig. 11. In the frequency domain, the SHC of the received NLFM, LFM-W80 and LFM-W10 signals were respectively -18.8 dB, -19.5 dB and -17.8 dB below the fundamental frequency component. The SNR of the SHC of NLFM was improved by 0.7 dB when compared with the SHC of the LFM-W80 signal of same energy level. However, the SNR of the SHC of NLFM was 1 dB below with the SHC of the LFM-W10 signal due to less energy.

In method II, all the received NLFM, LFM-W80 and LFM-W10 signals were processed using PI. After the PI process, the SNR of the SHC of NLFM, LFM-W80 and LFM-W10 signals was improved by 6 dB and the fundamental frequency components were suppressed by 37 dB. The PI process applied to the nonlinear received NLFM signals is shown in Fig. 12 whereas the PI process for the LFM signals is shown in Fig. 13. After the application of PI, the extracted second harmonic chirps of NLFM, LFM-W80 and LFM-W10 signals were processed using associated harmonic matched and mismatched filters to perform second harmonic pulse compression.

The envelopes of the compressed second harmonic chirps of NLFM, LFM-W80 and LFM-W10 signals after processed with the PI are shown in Fig. 14 and the pulse compression results are shown in Table 4. The results indicated that the PSL of the compressed second harmonic signal of the NLFM chirp was improved by 17.6 dB and 5 dB when compared with the PSL of LFM-

W80 and LFM-W10 signals respectively. Compared to the results in method-I, the PSL of the compressed second harmonic signal of the LFM-W10 was improved by 7.2 dB due to the application of PI. Similarly the ISL of the compressed second harmonic signal of the NLFM was improved by 13.5 dB and 10.8 dB when compared with the ISL of the LFM-W80 and LFM-W10 signals, respectively.

The MLW of the compressed second harmonic chirps of NLFM, LFM-W80 and LFM-W10 signals was reduced by more than 50% in method II due to the application of the 45% FBW excitation signals which therefore can improve the axial resolution. Similar to the method-I, the MLW of the compressed second harmonic chirp of the NLFM was higher due to the lower -6 dB bandwidth but comparable to the LFM-W80 and LFM-W10 signals. In method II, the pulse compression performance of the harmonic matched filter greatly relies on the PI. Therefore the MLW of the compressed second harmonic chirp will be further increased if the PI fails to extract the complete SHC under the tissue motion.

Doppler Sensitivity Evaluation of Designed Chirp Signals

The pulse compression performance of the harmonic matched filter will reduce when its frequency response is not spectrally matched with the SHC of the received signal. In the human body this occurs when the backscattered echoes are received from moving blood (or tissue) causing a Doppler shift in the signal spectrum. The amount of Doppler shift is directly proportional to the velocity of the moving blood and is inversely proportional to the speed of sound in the tissue. The Doppler shift can be expressed as Jensen (1996),

$$f_D = -\frac{2v}{c} f_c \tag{14}$$

where f_D is the Doppler shift in frequency, v is the velocity of the moving blood, c is the speed of sound in the tissue, and f_c is the centre frequency of the excitation signal.

In order to measure the Doppler tolerance of the NLFM and reference LFM signals, the Doppler shift frequency (f_D) was calculated and added to the centre frequency (f_c) of the harmonic matched filter. It was assumed that the velocity (v) of the moving blood is 2 m/s, the speed of sound (c) in the tissue is 1500 m/s, and the centre frequency (f_c) of the SHC is 4.5 MHz. A Doppler shift frequency (f_D) of 12 kHz was computed using (14) and added to the centre frequency of the harmonic matched filter to mimic the Doppler shift caused by the moving blood. The measured nonlinear signals from the hydrophone with a FBW of 20% were processed using associated harmonic matched and mismatched filters with an added Doppler shift frequency of 12 kHz. The compressed second harmonic chirp signals are shown in Fig. 15. It was found that all three chirp waveforms were robust to a Doppler shift of up to 2 m/s with no significant differences were observed in the MLW, PSL and ISL of the compressed signals when compared with the experimental results of method I shown in Table 4.

Discussion and Conclusion

In this study, NLFM and LFM could use long signals because a single element probe was employed in the experiments. Moreover, the long duration signal also increases the total excitation energy and thus improve the

SNR, which was required to generate the nonlinear harmonics in the water experiments. However, the long duration signal has a limitation imposed by the medical probes used in practical imaging systems (O'Donnell, 1992). Therefore, the duration of the chirp signals should be optimized for medical probes used in imaging applications.

In Method I, chirp signals with a fractional bandwidth (FBW) of 20% were proposed with a single transmission. The low bandwidth was used in order to avoid spectral overlapping between the required second harmonic and fundamental components in the nonlinear received signals. Therefore, a simple harmonic matched filter at the receiver was able to extract and compress the second harmonic component. The bandwidth of the SHC is twice the fundamental component, therefore the MLW of the SHC was improved compared to the MLW of the fundamental component. Moreover, the requirement of a single transmission of a chirp signal can avoid the motion artifacts and reduction of the system frame-rate as required by the pulse inversion process with the 45% FBW chirp signals in Method II. However, for imaging applications the MLW of the SHC is still so large, due to the use of 20% FBW, which degrades the axial resolution of the image.

In Method II, figures 7 and 11 show less spectral overlapping for NLFM and LFM-W80 chirps compared to the LFM-W10 signal due to the strong amplitude tapering used in the design process. Although all applied signals have same sweeping bandwidth, however, their -6 dB bandwidths were altered due to the application of different amplitude tapering. The -6 dB bandwidth of the NLFM chirp was respectively 40% and 11% lower than the LFM-W10 and LFM-W80 signals. In this method, the MLW of the compressed second harmonic chirp signals was improved by 50% due to the use of 45% FBW chirp signals which can improve the axial resolution. However, due to the use of PI for the extraction of SHC, the system frame-rate was reduced by half compared to the Method I.

The results of second harmonic pulse compression were also compared with the previous study of nonlinear quadratic chirp for tissue harmonic imaging (Song et al., 2010). It was observed that the PSL of the compressed second harmonic signal of the proposed NLFM chirp was improved by at least 5 dB in Method-I and at least 2 dB in Method II with the comparative axial MLW.

The nonlinear chirp was carefully designed with the customised window so that the signal spectrum should match with the frequency response of the transducer and maximum energy was transmitted through the transducer. At the receiving side, this results in improvement in the SNR of the SHC and reduction of the PSL after second harmonic pulse compression. Moreover, for the NLFM chirp, the second harmonic pulse compression was done without the need of an additional windowing on the harmonic matched filter, which can potentially avoid the reduction of axial resolution and the gain in the SNR.

The NLFM chirp and associated harmonic matched filter were also robust to the Doppler shift caused by the moving blood of up to 2 m/s and frequency dependent attenuation of 0.5 dB/[cm×MHz] up to penetration depth of 5 cm with no significant degradation of the MLW, PSL and ISL after pulse compression.

Acknowledgements

Authors are thankful to the Mehran University of Engineering & Technology, Jamshoro, Sindh, Pakistan, for providing all necessary facilities to carry out this research.

References

Adams JW. A new optimal window [signal processing]. IEEE Transactions on signal processing, 1991;39:1753–1769.

Anderson M. A 2D nonlinear wave propagation solver written in open-source matlab code. In: Ultrasonics Symposium (IUS), IEEE. 2000. pp. 1351 – 1354.

Arif M. Ultrasound harmonic imaging using chirp coded excitation. Ph.D. thesis, University of Leeds, 2010.

Arif M, Cowell D, Freear S. Pulse compression of harmonic chirp signals using the fractional fourier transform. Ultrasound in medicine & biology, 2010a;36:949–956.

Arif M, Harput S, Freear S. Performance evaluation of nonlinear frequency modulated signals in ultrasound harmonic imaging. In: Ultrasonics Symposium (IUS), IEEE, 2010b. pp. 2016–2019.

Arshadi R, Yu ACH, Cobbold RSC. Coded excitation methods for ultrasound harmonic imaging. Canadian Acoustics, 2007;35:35–46.

Averkiou M. Tissue harmonic imaging. In: Ultrasonics Symposium (IUS), IEEE. 2000. pp. 1563–1572.

Borsboom J, Chin CT, Bouakaz A, Versluis M, de Jong N. Harmonic chirp imaging method for ultrasound contrast agent. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2005;52:241–249.

Brandon P. Design of a nonlinear pulse-compression system to give a low-loss high-resolution radar performance. Marconi Review, 1973;36:1–45.

Chiao R, Hao X. Coded excitation for diagnostic ultrasound: a system developer's perspective. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2005;52:160–170.

Cobbold RSC. Foundations of Biomedical Ultrasound. Oxford University Press, New York, NY, 2007.

Collins T, Atkins P. Nonlinear frequency modulation chirps for active sonar. IEE Proceedings - Radar, Sonar and Navigation, 1999;146:312–316.

Cook C, Bernfeld M. Radar signals: An introduction to theory and application. New York: Academic Press, 1967.

de Jong N, Bouakaz A, Frinking P. Basic acoustic properties of microbubbles. Echocardiography, 2002;19:229–240.

Duck F. Tissue non-linearity. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, 2010;224:155–170.

Girod B, Rabenstein R, Stenger A. Signals and Systems. John Wiley & Sons, Ltd, Chichester, West Sussex, England, 2001.

Gran F, Jensen J. Designing waveforms for temporal encoding using a frequency sampling method. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2007;54:2070–2081.

Hallaj IM, Cleveland RO, Hynynen K. Simulations of the thermo-acoustic lens effect during focused ultrasound surgery. The Journal of the Acoustical Society of America, 2001;109:2245–2253.

Harput S, Arif M, Mclaughlan J, Cowell D, Freear S. The effect of amplitude modulation on subharmonic imaging with chirp excitation. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2013;60:2532–2544.

Harris F. On the use of windows for harmonic analysis with the discrete fourier transform. Proceedings of the IEEE, 1978;66:51–83.

Jensen JA. Estimation of Blood Velocities Using Ultrasound: A Signal Processing Approach. Cambridge University Press, New York, NY, 1996.

Jensen JA. Medical ultrasound imaging. Progress in Biophysics and Molecular Biology, 2007;93:153 – 165.

Johnston J, Fairhead A. Waveform design and doppler sensitivity analysis for nonlinear fm chirp pulses. IEE Proceedings Communications, Radar and Signal Processing, 1986;133:163–175.

Kim D, Lee J, Kwon S, Song T. Ultrasound second harmonic imaging with a weighted chirp signal. In: Ultrasonics Symposium (IUS), IEEE. 2001. pp. 1477–1480.

Ma Q, Ma Y, Gong X, Zhang D. Improvement of tissue harmonic imaging using the pulse-inversion technique. Ultrasound in Medicine & Biology, 2005;31:889 – 894.

Maresca D, Skachkov I, Renaud G, Jansen K, van Soest G, de Jong N, van der Steen AFW. Imaging microvasculature with contrast-enhanced ultraharmonic ultrasound. Ultrasound Med Biol, 2014;40:1318–1328.

Milleit R. A matched-filter pulse-compression system using a nonlinear fm waveform. Aerospace and Electronic Systems, IEEE Transactions on, 1970;AES-6:73–78.

Misaridis T, Jensen J. Use of modulated excitation signals in medical ultrasound. Part I: basic concepts and expected benefits. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2005a;52:177–191.

Misaridis T, Jensen J. Use of modulated excitation signals in medical ultrasound. Part II: design and performance for medical imaging applications. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 2005b;52:192–207.

Misaridis T, Pedersen M, Jensen J. Clinical use and evaluation of coded excitation in B-mode images. Ultrasonics Symposium (IUS), IEEE. 2000;2:1689–1693.

O'Donnell M. Coded excitation system for improving the penetration of real-time phased-array imaging systems. and Frequency Control IEEE Transactions on Ultrasonics, Ferroelectrics, 1992;39:341–351.

Pollakowski M, Ermert H. Chirp signal matching and signal power optimization in pulse-echo mode ultrasonic nondestructive testing. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 1994;41:655–659.

Ramalli A, Guidi F, Boni E, Tortoli P. A real-time chirp-coded imaging system with tissue attenuation compensation. Ultrasonics, 2015;60:65–75.

Rao N. Investigation of a pulse compression technique for medical ultrasound: a simulation study. Medical and Biological Engineering and Computing, 1994;32:181–188.

Simpson D, Chin CT, Burns P. Pulse inversion doppler: a new method for detecting nonlinear echoes from microbubble contrast agents. Ultrasonics, Ferroelectrics and Frequency Control, IEEE Transactions on, 1999;46:372–382.

Song J, Chang JH, Song Tk, Yoo Y. Coded tissue harmonic imaging with nonlinear chirp signals. Ultrasonics, 2011;51:516–521.

Song J, Kim S, Sohn HY, Song TK, Yoo YM. Coded excitation for ultrasound tissue harmonic imaging. Ultrasonics, 2010;50:613–619.

Wells PNT. Ultrasound imaging. Physics in Medicine and Biology, 2006;51:R83–R98.

Wojcik G, Mould J, Ayter S, Carcione L. A study of second harmonic generation by focused medical transducer pulses. In: Ultrasonics Symposium (IUS), IEEE. 1998. pp. 1583 –1588.

Figure Captions

Figure 1: Instantaneous frequency functions (Top), and the chirp-rate functions of the excitation signals of 45% fractional bandwidth (FBW) (Bottom).

Figure 2: Amplitude envelopes (Top), and the power spectra of the excitation signals of 45% FBW along with the measured transfer function (TF) of the transducer used in the experiments (Bottom).

Figure 3: Proposed system flow chart illustrating the signal processing chain for the pulse compression of the second harmonic component (SHC) using the harmonic matched filter. Method I (Top), and Method II (Bottom).

Figure 4: Schematic diagram of the experimental setup.

Figure 5: Power spectra of the simulated nonlinear received signals with a FBW of 20%.

Figure 6: The simulation results of method I show the envelopes of the compressed second harmonic chirp signals with a FBW of 20%.

Figure 7: Power spectra of the simulated nonlinear received signals with a FBW of 45%.

Figure 8: The simulation results of method II showing the envelopes of the compressed second harmonic chirp signals of 45% FBW after processed with the PI.

Figure 9: Power spectra of the measured nonlinear received signals with a FBW of 20%.

Figure 10: The experimental results of method I show the envelopes of the compressed second harmonic chirp signals with a FBW of 20%.

Figure 11: Power spectra of the measured nonlinear received signals with a FBW of 45%.

Figure 12: Illustration of the PI process applied to the experimental data of NLFM chirp with a FBW of 45%. Received Rx1 signal (Top-Left) and associated power spectrum (Top-Right), Received Rx2 signal with a phase difference of 180° (Middle-Left) and associated power spectrum (Middle-Right), and summation of Rx1 and Rx2 NLFM signals in time domain (Bottom-Left) and associated power spectrum showing the enhancement of the SHC and suppression of the fundamental and third harmonic components (Bottom-Right).

Figure 13: Illustration of the PI process applied to the experimental data of LFM-W80 (Left) and LFM-W10 (Right) chirps with a FBW of 45%. (Top) power spectrum of the received Rx1 signal, (Middle) power spectrum of the received Rx2 signal with a phase difference of 180°, (Bottom) power spectrum of the time domain summation of Rx1 and Rx2 LFM signals showing the enhancement of the SHC and suppression of the fundamental and third harmonic components.

Figure 14: The experimental results of method II showing the envelopes of the compressed second harmonic chirp signals of 45% FBW after

processed with the PI.

Figure 15: Envelopes of the compressed second harmonic chirp signals with a FBW of 20% processed by associated harmonic matched and mismatched filters with and without an added Doppler shift frequency of 12 kHz.

Tables

Table 1: Design parameters of excitation signals.

Excitation Signals		NLFM	LFM
Parameters	Symbols	Values	
Sampling Frequency	f_s	40 MS/s	40 MS/s
Centre Frequency	f_c	2.25 MHz	2.25 MHz
Fractional Bandwidth	Δf	20%, 45%	20%, 45%
Time Duration	T	20 μs	20 μs
Gamma	γ	1.20	
Alpha	α	0.40	

Table 2: Simulation parameters and their values.

Parameters	Symbols	Values
Peak pressure	p	1 MPa
Axial distance	z	5 cm
Speed of sound	c	1500 m/s
Material density	ρ	1000 kg/m
Nonlinear coefficient	B/A	5.2
Frequency dependent attenuation	α_o	0.5 dB/[MHz·cm]

Table 3: Simulation results showing the performance evaluation parameters of second harmonic pulse compression.

	Method I			Method II		
Compressed Chirp Signals	MLW (μs)	PSL (dB)	ISL (dB)	MLW (μs)	PSL (dB)	ISL (dB)
NLFM	5.5	-42.6	-43.7	2.4	-48.8	-47.8
LFM-W80	4.6	-26.9	-34.7	2.0	-21.2	-34.2
LFM-W10	5.0	-26.5	-29.8	2.2	-33.8	-35.0

Table 4: Experimental results showing the performance evaluation parameters of second harmonic pulse compression.

	Method I			Method II		
Compressed Chirp Signals	MLW (μs)	PSL (dB)	ISL (dB)	MLW (μs)	PSL (dB)	ISL (dB)
NLFM	5.4	-41.2	-42.9	2.4	-38.2	-44.2
LFM-W80	4.5	-28.1	-34.5	2.0	-20.6	-30.7
LFM-W10	5.1	-25.6	-24.3	2.3	-33.2	-33.4