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Research Paper

SNR improvement and range side lobe suppression in Golay-encoded Doppler detection for ultrasound high-frequency swept-scan imaging system

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ABSTRACT

Background: Golay excitation can be utilized to improve the signal-to-noise ratio (SNR). However, its application in ultrasound Doppler detection is limited due to potential motion artifacts in conventional decoding process. In this study, the improved Golay decoding has been implemented in a commercial 40-MHz pre-clinical system and experimentally verified for its efficacy in color-flow imaging.
Methods: Based on the Doppler spectral difference of half pulse-repetition-frequency (*PRF*) between the main lobe and range side lobe components of Golay excitation, the slow-time decoding with a low-pass filter with cut-off frequency of *PRF*/4 can separate the desired main lobe from the side lobe interference. The Prospect® system has been customized to generate the Golay transmit sequence in swept-scan mode and to perform the *PRF*/4 decoding for *in-vivo* evaluation of mouse blood flow.
Results: Results indicate that the *PRF*/4 decoding can effectively eliminate the range side lobe artifacts in phantom experiments for flow velocity not exceeding the limit of ±*PRF*/4 in Doppler frequency. For *in-vivo* imaging, the SNR improves by about 7 dB and the Doppler penetration increases from 13.5 mm to 14.2 mm in mouse kidney when the transmit is switched from un-coded to Golay excitation. In mouse abdominal aorta, the Golay transmit also increases the Doppler sensitivity.
Conclusion: Golay-encoded color-flow imaging has been established using the pre-clinical system to achieve SNR improvement in Doppler detection without suffering from the range side lobe artifacts. To guarantee the performance of *PRF*/4 decoding, the side lobe aliasing should be avoided by carefully selecting the *PRF* to the match the flow velocity.

1. Introduction

Ultrasound high-frequency (>20 MHz) imaging system is capable of portraying fine anatomy with high image resolution and has exhibited its potential in applications of small animal pre-clinical imaging, dermatology and ophthalmology [1–6]. Moreover, the detection of microcirculation also improves in these applications because of the elevation of blood signal magnitude at high frequency [7,8]. Note that, in high-frequency imaging, the corresponding array transducer is sometimes unavailable due to the challenging fabrication of small-pitch array elements with adequate electrical and acoustical isolation. On the contrary, in the swept-scan system [9], only one single-element transducer is utilized for lateral scanning to continuously acquire neighboring scan lines over the region of interest during the transducer movement. Due to its simplicity and low cost, the swept-scan high-frequency imaging system remains popular and commercially available such as VisualSonics Vevo 770® and S-Sharp Prospect®. For color-flow (CF) imaging in the swept-scan system, the scan line density will increases so that the sample volume can be successively interrogated by a group of overlapping scan lines. Then, slow-time Doppler processing can be directly performed in the lateral direction by treating each scan line as one Doppler sample. Therefore, even when the frame rate in CF imaging remains unchanged, the number of Doppler samples increases with the density of lateral scan lines and thus subsequent Doppler processing would benefit from the increased number of Doppler samples. However, due to the severe tissue attenuation and the fixed focal depth of single-element

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transducer, the penetration of high-frequency CF imaging in the swept-scan system is limited and thus the microvessel morphology beneath the focal depth suffers from low Doppler sensitivity.

Coded excitation can boost acoustic transmit energy without excessive pressure amplitude by using elongated waveform [10–12]. In receiving, the code energy can be compressed into a short time interval by matched filtering to restore the image resolution in the axial direction. The resultant improvement in signal-to-noise ratio (SNR) has been reported in synthetic aperture imaging [13–16], harmonic imaging [17–19], elasticity imaging [20] and Doppler detection [21–23]. Generally speaking, coded excitation can be designed using either frequency modulation (FM) or phase modulation (PM) of the transmit waveform. The well-known FM method is to sweep the instantaneous frequency such as the chirp waveform. For PM, the binary change of phase in the bit sequence is utilized such as in the Barker code and the Golay code. Compared to the chirp waveform, the PM waveform is less demanding in transmit hardware and provides higher SNR improvement [12]. Among these PM codes, the Golay code can achieve complete elimination of range side lobes but at the cost of a pair of complementary transmits for each scan line. For example, the 2-bit Golay pair is composed of two bit sequences [−1 −1] and [1 −1]. For decoding, the received echo is respectively compressed by matched filtering before the summation of filter output to eliminate the range side lobe as shown in Fig. 1. Generally, a pair of L-bit binary sequences $A(n)$ ($n = 1 \ldots L$) and $B(n)$ ($n = 1 \ldots L$) is a complementary Golay pair when

$$A(n) * A(-n) + B(n) * B(-n) = 2L\delta(n) \tag{1}$$

In (1), the symbol $*$ represents the convolution in matched filtering and $\delta(n)$ is the delta function. For better transmit efficiency, the Golay code will also be up-sampled and convolved with a radio-frequency waveform whose center frequency matches the pass band of transducer [24].

For swept-scan CF imaging, however, the complete cancellation of Golay range side lobes in (1) is not possible due to the movement of blood flow (*i.e.*, the imaged object) and the scanning of transducer (*i.e.*, the imaging system itself). This motion compromises the decoding process and blurs the CF image due to the presence of range side lobe artifacts. Consequently, Golay-encoded Doppler detection is conventionally limited to low-velocity flow or quasi-stationary tissue in ultrasound imaging [25]. Otherwise, additional computation may be required to provide motion compensation for Golay decoding [26]. This is why the coded excitation in high-frequency imaging is generally the FM chirp waveform [27,28].

Alternative solution to suppress the range side lobe for moving objects is weighted Golay sum [29] in which the received echoes from three successive Golay transmits as [A, B, A] can be respectively weighted by [0.5 1 0.5] before the summation. Noted that the weighted Golay sum is equivalent to a slow-time decoding filtering across the successive Golay transmits by using the weighting value as the filter coefficients and thus can be readily integrated in the Golay-encoded Doppler detection. This is particularly true for the swept-scan system because the large number of Doppler samples would facilitate the design and implementation of Golay decoding filter in the slow-time domain. Nevertheless, to the authors' knowledge, the Golay decoding filter in the literatures remains to be empirically determined.

In our previous study, a theoretical guideline for optimal filter design has been established to suppress the range side lobe artifacts in the slow-time domain [30]. With a given pulse repetition frequency (*PRF*), the proposed *PRF*/4 slow-time processing has been preliminarily verified for Golay-encoded Doppler detection to remove the range side lobe artifacts. In this paper, Golay-encoded Doppler is implemented in a small animal pre-clinical imaging system with the proposed *PRF*/4 slow-time processing to examine its efficacy in *in-vivo* mouse imaging. First, the derivation of the *PRF*/4 Golay decoding filter is briefly reviewed in terms of the Doppler spectral characteristics of the main lobe and the side lobe components in Section 2. In Section 3, the experimental setup used to verify the *PRF*/4 Golay decoding in the pre-clinical imaging system is provided. The achievable SNR improvement and the corresponding CF images with Golay excitation are presented in Section 4. Discussions and conclusions are provided in Section 5.

2. The *PRF*/4 Golay decoding

In conventional Doppler detection, the same transmit waveform is repetitively fired to the imaged object and the change of distance between the ultrasound transducer and the imaged object is detected by calculating the phase shift among echoes from consecutive firings. In other words, the phase shift in the slow-time direction contains essential information about the moving speed of imaged object. In Golay-encoded Doppler, however, the transmit waveform changes from A to B and vice versa in consecutive firings. Based on the complementary Golay property, the A transmit and the B transmit have to be respectively decoded so that the required flow information can be recovered in the main lobe component. Nonetheless, the side lobe component is also present after decoding and it will interfere with the desired flow information. In the following, the *PRF*/4 Golay decoding is explained in details about how to separate the main lobe component from the side lobe component in Doppler detection.

In Golay-encoded Doppler, the Golay pair in (1) is repeated for transmit in successive scan lines as [$A(1)$, $B(1)$, $A(2)$, $B(2) \ldots$, $A(N)$, $B(N)$]. Assuming the imaged object is moving toward the ultrasound transducer at a constant speed v and the received echo from each transmission has been compressed by matched filtering in the fast-time domain, these compressed echoes can be arranged into a two-dimensional matrix $E(k, t)$. Note that each row in $E(k, t)$ represents the echo from the k-th transmission and thus the k and t are the slow-time index and fast-time index, respectively. Due to the motion of imaged target, both the main lobe component in blue color and the side lobe component in yellow color experience the same time delay of $\Delta\tau_d$ between adjacent rows as shown in Fig. 2. Nonetheless, the main lobe component keeps the same polarity while the side lobe component inverts its polarity between adjacent rows. Consequently, in order to account for the polarity change and the time delay due to motion, $E(k, t)$ can be formulated as

$$E(k,t) = \begin{bmatrix} +S(t) & +M(t) & +S(t) \\ -S(t-1\Delta\tau_d) & +M(t-1\Delta\tau_d) & -S(t-1\Delta\tau_d) \\ +S(t-2\Delta\tau_d) & +M(t-2\Delta\tau_d) & +S(t-2\Delta\tau_d) \\ \vdots & \vdots & \vdots \\ \pm S(t-k\Delta\tau_d) & +M(t-k\Delta\tau_d) & \pm S(t-k\Delta\tau_d) \\ \vdots & \vdots & \vdots \\ \vdots & \vdots & \vdots \end{bmatrix} \tag{2}$$

In (2), the main lobe component from the k-th transmission is denoted by $+M(t-k\Delta\tau_d)$ and the corresponding side lobe component is $\pm S(t-k\Delta\tau_d)$ where the polarity depends on the value of k. Compared to that of the main lobe component, the phase shift of the side lobe component differs by an additional π between adjacent rows as shown in the following:

$$\begin{aligned} \Delta\phi_M &= 2\pi f_0 \Delta\tau_d \\ \Delta\phi_S &= 2\pi f_0 \Delta\tau_d + \pi \end{aligned} \tag{3}$$