Biomedical Signal Processing and Control 13 (2014) 306–312

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

Barker coded excitation with linear frequency modulated carrier for ultrasonic imaging

Juan Fu, Gang Wei, Qinghua Huang*, Fei Ji, Yizhi Feng

School of Electronic and Information Engineering, South China University of Technology, Guangzhou, China

ARTICLE INFO

Article history:
Received 12 February 2014
Received in revised form 26 May 2014
Accepted 8 June 2014
Available online 28 June 2014

Keywords:
Ultrasound imaging
Barker coded excitation
LFM carrier
Pulse compression
Axial resolution and SNR improvements

ABSTRACT

A new Barker coded excitation using linear frequency modulated (LFM) carrier (called LFM-Barker) is proposed for improving ultrasound imaging quality in terms of axial resolution and signal-to-noise ratio (SNR). The LFM-Barker coded excitation has two independent parameters: one is the bandwidth of LFM carrier, and the other is the chip duration of Barker code. To improve the axial resolution, increase the bandwidth of LFM carrier; and to improve the SNR, increase the chip duration of Barker code. In this study, a LFM pulse with proper (<5.5) time–bandwidth product is considered as the carrier in order to avoid sidelobes inside the mainlobe of matched filtered output. A pulse compression scheme for the LFM-Barker coded excitation is developed, and it consists of the LFM matched filter and Barker code mismatched filter. In the simulations, the impulse response of transducer can be approximated by a Gaussian shaped sinusoid with 5 MHz central frequency of 60% −6 dB fractional bandwidth. The pulse compression filter is performed to suppress sidelobes below −40 dB roughly, which is acceptable in medical imaging. Simulation results show that in comparison with conventional Barker coded excitation using sinusoid carrier (called Sinusoid-Barker), the axial resolution of the LFM-Barker coded excitation system can be doubled, and the SNR can be improved by about 3 dB. Simulation of B-mode images with the Field II program demonstrates that the axial resolution is improved from 0.7 mm to 0.4 mm. In addition, the LFM-Barker coded excitation is robust for frequency dependent attenuation of tissues.

1. Introduction

Ultrasound imaging is an important medical imaging modality, and is often used for many diagnostic purposes, mostly considering that it is safe and produces real-time images [1], moreover less expensive and simpler in use than other imaging techniques [2,3]. Axial resolution and signal-to-noise ratio (SNR or penetration) are crucial factors to consider in evaluation of the image quality of ultrasound system. It is well known that the axial resolution can be improved by increasing the frequency of the transmitted ultrasound. Unfortunately, the energy of ultrasound is attenuated when propagating through tissues. Generally the attenuation of ultrasound in tissues is frequency dependent, and increases as the frequency increases. Therefore, the improvement in the axial resolution results in limited penetration depth and a decrease in SNR of the received signals due to the higher attenuation coefficients.

On the other side, the improvement in SNR can be achieved by increasing the total ultrasound energy input into the system, by increasing either the pulse amplitude or pulse duration. In medical ultrasound, safety requirements limit the peak amplitude of the pulse in order to avoid heating and cavitation damages within the insonated human body [4,5]. Therefore, the way to improve the SNR is to excite the transducer with a long and modulated pulse (coded excitation) in the transmitter, which allows increasing the average transmit power without increasing the peak amplitude of the pulse. Moreover, in the receiver, the echoes of the long pulses can be compressed (decoded) with pulse compression filter to restore the axial resolution.

Coded excitation technique is often used in medical ultrasound for improving the SNR and increasing the penetration depth. It has been studied and applied to medical ultrasound for almost 10 years [6–14]. A variety of coded excitations have been developed, including the linear frequency modulated (LFM) signal or nonlinear frequency modulated signal [6–8], and the phase coded signal, such as Barker codes [9,10], m-sequences [11] and Golay complementary sequences [12], and the amplitude modulated signal [13,14]. They can be basically categorized according to their modulation functions into three types: frequency modulation, phase modulation and amplitude modulation. Generally, a coded waveform uses only one of modulation functions. For example, Barker coded signal uses

* Corresponding author. Tel.: +86 20 87113540.
E-mail address: qhhuang@scut.edu.cn (Q. Huang).

http://dx.doi.org/10.1016/j.bspc.2014.06.004
1746-8094/

the bi-phase modulation, and the LFM coded signal uses the linear frequency modulation.

In this paper, we focus on improving the performance of Barker coded excitation. Conventionally Barker code is modulated with a single-frequency carrier of sinusoidal wave, and the Barker coded excitation using the sinusoid carrier (called Sinusoid-Barker) has the advantages of simpler pulser and lower cost compared with the LFM coded excitation [15]. However, its frequency spectrum cannot accord well with ultrasound transducer frequency spectrum. Moreover, the single-frequency carrier of sinusoidal wave has the time–bandwidth (*TB*) product on the order of one, and hence the gain in SNR (SNRG) through the Sinusoid-Barker coded excitation can be only achieved by Barker code length [16]. Theoretically, Barker coded waveforms with longer code length give better SNRG. Unfortunately, the Barker-13 (i.e. the code length is 13) is the longest Barker code available, and the Barker-13 coded excitation can achieve 11.1 dB SNRG [17]. Therefore the SNRG is limited for the Sinusoid-Barker coded excitation. In addition, the Sinusoid-Barker coded signal has frequency-shift sensitivity [9]. Because of frequency dependent attenuation of tissues, it cannot be well applied in medical ultrasound.

Due to the drawbacks of conventional Sinusoid-Barker coded excitation, a new Barker coded excitation using the LFM pulse as the carrier (called LFM-Barker) is proposed in this paper. The LFM-Barker coded signal uses two types of modulation functions. Within every chip of Barker code, it uses the linear frequency modulation. Between chips of Barker code, it uses the Barker phase modulation [18]. The LFM pulse as the carrier is flexible to control its frequency spectrum according to the impulse response of ultrasound transducer, and it has the *TB* product on the order of above one [7]. Hence, the SNRG through the LFM-Barker coded excitation can be achieved not only by the Barker code length but also by the *TB* product of LFM carrier. Moreover, the LFM signal has the advantage of frequency-shift tolerance [7].

However, the major problem associated with coded excitation is the range sidelobe artifacts [7–10,19]. In the Barker-13 coded excitation system, the peak sidelobe level (PSL) of matched filtered output is −22.3 dB below the mainlobe, which is not tolerated in medical ultrasound imaging. In this study, the mismatched filter for the LFM-Barker coded excitation is developed to decode the received echo and suppress the sidelobe level.

In this paper, basic concepts and expected benefits of the LFM-Barker coded excitation are presented. The two Barker coded excitations using the LFM carrier and sinusoid carrier, respectively, are compared in simulation experiments. Section 2 gives the representations of the LFM-Barker coded excitation and introduces the mismatched filter for pulse compression. The optimized coded excitation/pulse compression scheme was simulated to verify the quality of ultrasound imaging in Section 3. Section 4 draws the final conclusions.

2. Methods

2.1. LFM-Barker coded excitation

The ultrasound imaging system can be mathematically expressed as

$$e(t) = s(t) * H(t) + n(t) \tag{1}$$

where $*$ is the convolution operator, $s(t)$ is the excitation pulse, $H(t)$ is the system transfer function, $e(t)$ is the received echo and $n(t)$ is the additive noise [15].

In the Barker coded excitation system, Barker code is always modulated with a carrier. The encoding process can be described as a convolution of a carrier sequence with an oversampled Barker code sequence [9,12]. The Barker coded signal can be expressed by

$$s(n) = v(n) * c(n) \tag{2}$$

where $v(n)$ is the sample sequence of the carrier denoted as $v(t)$, and $c(n)$ is the oversampled Barker code sequence, as expressed by

$$c(n) = \sum_{k=0}^{P-1} c_k \delta(n - kT_p f_s) \tag{3}$$

where $\{c_k = \pm 1,\ k = 0,\ 1,\ \ldots,\ P-1\}$ is the original Barker code sequence before oversampling, P is the Barker code length, T_p is the chip duration of Barker code and f_s is the system sampling rate. The total duration of the Barker coded signal is $T = PT_p$.

Generally one or multiple cycles of sinusoidal wave at the transducer central frequency is used as the carrier of conventional Sinusoid-Barker coded excitation, and thus $v(t)$ can be expressed as

$$v_1(t) = \sin(2\pi f_0 t), \quad t \subset [0, T_p] \tag{4}$$

where f_0 is the transducer central frequency. The sinusoid carrier is a broadband signal containing all frequencies in a bandwidth of $1/T_p$ around f_0 [6,16]. It can be seen that the bandwidth of sinusoid carrier is determined by the chip duration of Barker code.

In this study, the carrier of the LFM-Barker coded excitation is formed by the LFM pulse with most energy concentrated within the transducer bandwidth, and thus $v(t)$ is expressed as

$$v_2(t) = \sin\left(2\pi\left(f_0 - \frac{B}{2}\right)t + \pi\mu t^2\right), \quad t \subset [0, T_p] \tag{5}$$

where B is the bandwidth of LFM carrier, and μ is the frequency sweep rate, which is $\mu = B/T_p$ corresponding to the chip duration of Barker code T_p and the bandwidth of LFM carrier B.

Generally the axial resolution of ultrasound imaging system is related to the bandwidth of transducer frequency response. Therefore, the bandwidth of the excitation signal should be as large as that of the transducer, resulting in the echoes with enlarged bandwidth, thereby improving the axial resolution. The bandwidth of Barker coded signal is basically determined by its carrier frequency spectrum [9]. In this study, the bandwidth of Barker coded signal must be redefined as the bandwidth of carrier, since the definition of bandwidth for a phase encoded signal is complex, due to the sharp phase transitions in the signal [16]. The sinusoid carrier has the *TB* product on the order of one. When the bandwidth of sinusoid carrier is increased to improve the axial resolution, the chip duration of Barker code is decreased and the SNR is deteriorated for the Sinusoid-Barker coded excitation.

In contrast, the bandwidth of LFM carrier can be flexibly chosen regardless of the chip duration of Barker code because the LFM carrier has the *TB* product on the order of above one. It is feasible to broaden the bandwidth of the LFM-Barker coded signal in order to match the transducer bandwidth, thereby achieving the improvement in the axial resolution.

2.2. Pulse compression scheme

In an ultrasound coded excitation system, the returned echo is squeezed (decoded) into a short (compressed) pulse through pulse compression filter in the receiver [12]. Thus, the compressed (decoded) signal $d(t)$ is

$$d(t) = e(t) * p(t) \tag{6}$$

where $p(t)$ is the pulse compression filter.

If the matched filter (the time inverse of the transmitted signal) is used for pulse compression, the output is an autocorrelation function (ACF) of the transmitted signal. The ACF envelopes