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Limitation of coded excitation for synthetic aperture focusing in measuring fast-moving tissues

移動する組織の測定に対する符号化送信による開口合成の限界

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1. Introduction

For blood flow estimation a conventional ultrasound imager calculates a 2-D image of the axial flow velocity along the beam axis [1]. This imager has two limitations: one is its low temporal resolution, and the other is that it calculates only the 1-D axial velocity. Several groups utilized synthetic aperture focusing with coded excitation to improve temporal resolution [2-4]. In this method, all elements transmit pulses with temporal coding. The employment of temporal coding acquires the focused transmit beam when the target velocity is negligible compared to the sound velocity. However, this assumption may be inappropriate in measuring fast-moving tissues, e.g. blood flow estimation and heart wall measurement. In this study, we investigate the deterioration of the performance of the ultrasound imaging method using coded exicitation that employs synthetic aperture focusing in measuring fast-moving tissues.

2. Blurred Image Acquired by a Synthetic Aperture Imager with Coded Excitation

Since a conventional delay-and-sum beamformer employs a set of time delays at each transmit event, all the transmit pulses arrives the measurement point at the same time. In contrast, a synthetic aperture imaging method with coded excitation fixes a set of time delays at the transmit event. Therefore, the variation of time-of-flight among transmit elements causes the arrival-time difference of transmit pulses at the measurement point, as shown in Fig. 1. The employment of long pulses in coded excitation also increases the arrival-time variation. Since the arrival-time variation of transmit pulses and target velocity vary the location at the single point target, a blurred image of fast-moving tissue may be acquired by a synthetic aperture imager with coded excitation.

In this study, we assume that all elements of a linear array are excited at the same time. The

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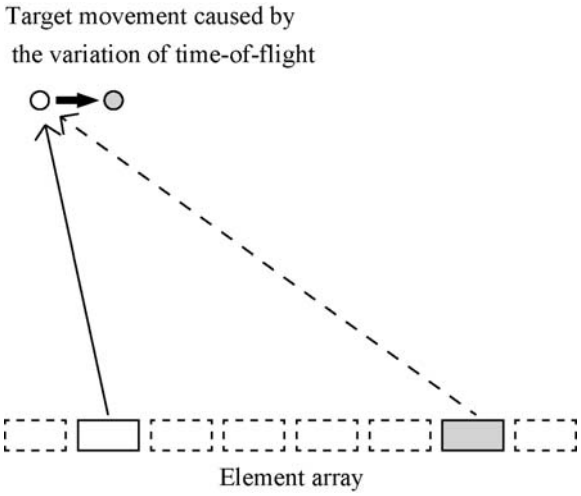

Fig. 1 Target movement caused by the arrival-time difference of transmit pulses, blurring the image acquired by a synthetic aperture imager with coded excitation.

reflection point of a pulse transmitted at a k-th element, Q_k, satisfies the following equations:

$$|P_k Q_k| = ct_k, \tag{1}$$

$$\mathbf{p}_k = \mathbf{p}_0 + \mathbf{v}t_k, \tag{2}$$

where $\mathbf{p}_0$ is the position vector of a point target at the excitation time, $\mathbf{p}_k$ is the position vector of the reflection point P_k, c is the sound velocity, $\mathbf{v}$ is the velocity vector of the point target, t_k is the arrival time of the k-th pulse at the point target. The received signal after delay-and-sum process is expressed as follows:

$$s(t) = \sum_{k,l} s_P(t - T_{k,l} + T_{Mk,l}), \tag{3}$$

$$T_{k,l} = (|Q_k P_k| + |P_k Q_l|)/c, \tag{4}$$

$$T_{Mk,l} = (|Q_k M| + |M Q_l|)/c, \tag{5}$$

,where Q_l is a l-th receive element, M is a measurement point and $s_P(t)$ is the signal returned from a point target after pulse compression.

In a numerical analysis, we investigate the variation of the location reflected at a point target and its effect on the received signal after a delay-and-sum process. We use a 25 mm linear array with the element pitch of 0.2 mm. The tissue velocity is 10 m/s in the lateral direction. We neglect the effect of the pulse length on the arrival-time variation, and we assume that the signal after pulse compression, $s_P(t)$, becomes an ultrasound pulse, where its center frequency is 7.5 MHz and the –6 dB bandwidth is 4.5 MHz.

3. Results

Fig. 2 shows the distribution of locations reflected at a single point target. Since the variation of pass-length increases in measuring a shorter range, the reflection points at the depth of 10 mm spread over a wide region.

The wave length of a 7.5 MHz ultrasound pulse is 0.2 mm. The simulation result indicates that the 10-m/s tissue velocity distributes the reflection points over 0.25- and 0.5-λ regions at the depth of 10 and 30 mm, respectively. The distribution of the reflection points causes the time-shift error of 0.5- and 1-λ at the maximum in a delay-and-sum process. The time-shift error caused by the arrival-time difference brings the waveforms of received signals after the delay-and-sum process, as shown in Fig. 3. 2-D speckle tracking is used for the estimation of the tissue 2-D vector velocity, where the 2-D speckle tracking calculates the 2-D cross correlation of the tissue speckle [5]. Since the waveform change of the received signal causes unexpected change in tissue speckle, a synthetic aperture imager with coded excitation is supposed to be unsuitable to measure the 2-D vector velocity of a fast-moving tissue.

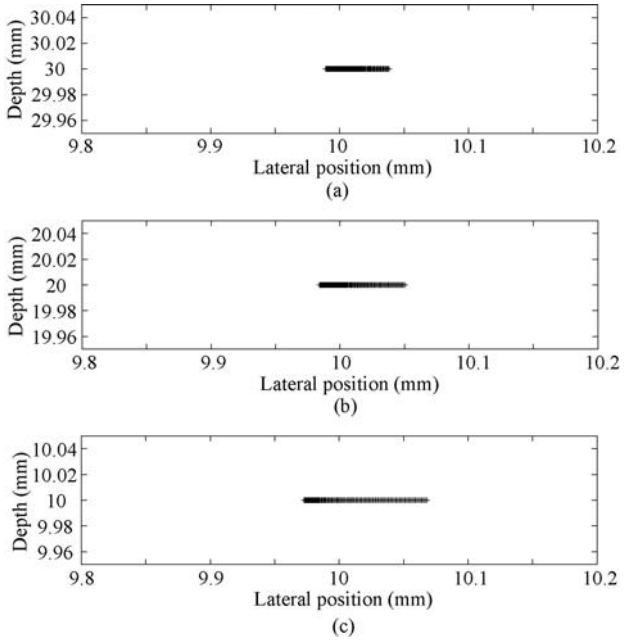

Fig. 2 Distribution of locations reflected at a single pint target located at the depths of (a) 30, (b) 20 and (c) 10 mm. Target velocity is 10 m/s in the lateral direction.

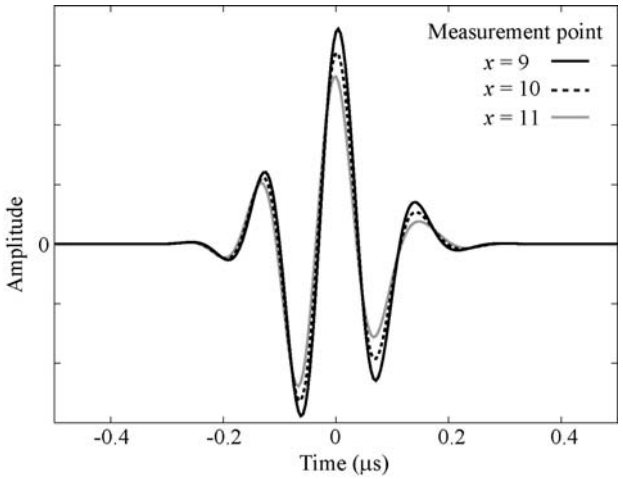

Fig. 3 Received signals after the delay-and-sum process, where the tissue velocity is 10 m/s in the lateral direction. Measurement points locate at the depth of 10 mm, and their lateral positions are 9, 10 and 11 mm.

Acknowledgment

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