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● *Original Contribution*

ERROR ANALYSIS OF ULTRASONIC TISSUE DOPPLER VELOCITY ESTIMATION TECHNIQUES FOR QUANTIFICATION OF VELOCITY AND STRAIN

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Abstract—Recent work in the field of Doppler tissue imaging has focused mainly on the quantification of results involving the use of techniques of strain and strain-rate imaging. These results are based on measuring a velocity gradient between two points, a known distance apart, in the region-of-interest. Although many recent publications have demonstrated the potential of this technique in clinical terms, the method still suffers from low repeatability. The work presented here demonstrates, through the use of a rotating phantom arrangement and a custom developed single element ultrasound system, that this is a consequence of the fundamental accuracy of the technique used to estimate the original velocities. Results are presented comparing the performance of the conventional Kasai autocorrelation velocity estimator with those obtained using time domain cross-correlation and the complex cross-correlation model based estimator. The results demonstrate that the complex cross-correlation model based technique is able to offer lower standard deviations of the velocity gradient estimations compared with the Kasai algorithm. (E-mail: mjb@ee.ed.ac.uk)

Key Words: **Doppler tissue imaging, Strain imaging, Strain rate, Velocity estimation, Kasai autocorrelation, Complex cross correlation model.**

INTRODUCTION

Doppler tissue imaging (DTI) was original described by McDicken et al. (1992) when it was demonstrated that a machine designed to perform blood flow velocity measurements could be used to measure the motion within the myocardium. Tissue Doppler based methods are now found on virtually all echocardiography machines, but the results would be improved by greater accuracy in the velocity measurement.

Recently, the technique of Doppler tissue imaging has been extended into the field of strain and strain rate imaging (Heimdal et al., 1998). These modalities are based on the principle of measuring the velocity gradient between two points and relating this to the mechanical strain of the material between the two points. Some recent work has demonstrated that certain conditions may be identified more clearly using this technique compared with standard Doppler tissue imaging: see, for example, Tsutsui et al. (1998); Hoffmann et al. (2002); Pislaru et al. (2002).

However, all of the estimations of the strain or strain rate are made from velocity estimations derived from DTI techniques; therefore, there is an obvious danger of compounding errors which occured in the original estimation. Despite the accuracy of DTI being quoted as around 10% to 20% (Fleming et al., 1994), there seem to be few publications looking at the accuracy of DTI techniques with a view to using the results for further estimations. Of course, many of the factors which degrade the velocity estimations made by DTI, such as the Doppler angle and overall cardiac motion, are cancelled out by the process of forming a velocity gradient. However, these are global artifacts and local artifacts such as errors due to noise in the signal or other unexpected artifacts will not cancel out. Kowalski et al. (2003) describe the potential value of ultrasonic regional strain analysis during dobutamine stress echocardiography. They conclude that the strain measurements allowed clear differentiation of three differing regional ischaemic

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substrates (ischaemic, stunned and scarred), although the results also demonstrated a suboptimal reproducibility (Garcia, 2003). The guest editorial in the same issue of the European Journal of Echocardiography, by Garcia (2003), describes how this lack of repeatability may be due to both the relatively low frame rate caused by the wide-sector scanning using in the work by Kowalski et al. (2003), as well as the low signal-to-noise ratio (SNR) when the difference between the two velocities being used to estimate the gradient is small compared with the accuracy of the original velocity estimation. In practice, these problems are partly overcome through the use of temporal and spatial averaging of the strain measurements. A single strain measurement is made using the average of several pairs of velocities, or a form of linear regression as described by Jackson and Thomas (2003). In the case of myocardial signals, the temporal averaging might be performed by averaging the strain over as many as three cardiac cycles (Dhooge, 2005). Brands et al. (1997) have relatively recently published work describing the use of a complex cross-correlation model (C3M) based velocity estimator, which is able to offer improved performance in terms of velocity estimation accuracy compared with time domain cross-correlation (XCorr) or Kasai autocorrelation (Kasai) techniques.

It is common for clinical scanners to estimate the velocity using some form of the Kasai autocorrelation algorithm (Kasai et al., 1985), which was originally developed to work with quadrature demodulated signals. These are formed by combining the received narrow-band signal with a copy of the transmitted signal. This produces an output consisting of both the sum of the transmitted and received frequencies and the required difference, or beat, frequency. This difference frequency is equal to the Doppler frequency induced by the motion of the scattering target. Filters are then applied to separate the Doppler frequency from the unwanted frequency components (Evans and McDicken, 2000). Pulse-wave Doppler devices work by estimating the frequency spectrum of the quadrature demodulated signal; however, the number of samples of the received signal required to obtain a sufficient frequency resolution makes this technique unsuitable for forming 2D images.

The algorithm developed by Kasai et al. (1985) uses multiple transmit/receive cycles and then calculates the autocorrelation across all of the received signals at a fixed depth. In this way, it is possible to build up a 2D image by estimating the velocity at a number of depths over a number of different directions. Considering the set of received signals for a single direction, the mean angular frequency will equate to the Doppler shift frequency. The mean angular frequency, $\bar{\omega}$, can be derived from the rate of change of phase, ϕ, using

$$\bar{\omega} = \phi'(0) = \frac{\phi(T)}{T} \tag{1}$$

where T is the time between each transmit/receive cycle, the reciprocal of which is known as the pulse repetition frequency. If the received signal is denoted by $z(t)$, then the autocorrelation, $R(T, t)$, may be determined using

$$R(T, t) = \int_{t-nT}^{t} z_1(t')\mathrm{d}t' \tag{2}$$

where n is the number of transmit/receive cycles used, t is the time it has taken for the received signal to return to the transducer and $z_1(t) = z(t)z^*(t - T)$, $z^*(t - T) = x(t - T) - jy(t - T)$. Here, $x(t)$ and $y(t)$ represent the in-phase and quadrature-phase components, respectively. The phase, ϕ, can then be determined as the argument of $R(T, t)$ as in eqn 3:

$$\phi(T, t) = \tan^{-1}\frac{R_y(T, t)}{R_x(T, t)} \tag{3}$$

It is clear from this that the output of the autocorrelation is a function of time and depends on the integration duration, nT, which is determined by the pulse repetition frequency and the number of transmit/receive cycles. Increasing the integration duration will improve the correlation, but will reduce the maximum frame rate achievable by the system.

The time domain cross-correlation method works by determining the shift in the position of the maximum value in the cross-correlation function of two received signals captured from separate transmit/receive cycles in a similar manner as for the Kasai method. The extent of this shift will depend on how far the scatterers have moved in the time between the transmit/receive cycles. Consider two received signals, $s_1(t)$ and $s_2(t)$: the cross-correlation may be expressed as in eqn 4:

$$R(\tau) = \int s_1(t)s_2(t+\tau)\mathrm{d}\tau \tag{4}$$

Assuming a sufficiently high pulse repetition frequency, then the two signals will be almost identical, except for one being a delayed version of the other; hence:

$$s_2(t) = s_1(t - t_s) \tag{5}$$

where t_s is the time lag between the two signals. Putting this back into eqn 4 gives:

$$R(\tau) = \int s_1(t)s_2(t - t_s + \tau)\mathrm{d}\tau = R_{11}(\tau - t_s) \tag{6}$$

where $R_{11}(t)$ is the autocorrelation function of $s_1(t)$ which will have a maximum when $\tau = t_s$. Therefore, the cross-