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New Doppler-based imaging method in echocardiography with applications in blood/tissue segmentation

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ABSTRACT

A parametric model for the ultrasound signals from blood and tissue is developed and a new imaging method, knowledge-based imaging, is defined. This method utilizes the likelihood ratio function to classify blood and tissue signals. The method separates blood and tissue signals by the difference in movement patterns in addition to the difference in powers. The prior information about the levels of expected system white noise and clutter noise are utilized to enhance the image quality. The implementation of knowledge-based imaging is outlined, and some knowledge-based images with different parameter settings are visually compared with a second-harmonic image, a fundamental image and a bandwidth image. In order to understand the parameter estimation process a computer simulation is introduced to outline the differences between the imaging methods. The apparent error rates are calculated in any reasonable tissue to blood signal ratio, tissue to white noise ratio and clutter to white noise ratio. A discussion of further development of knowledge-based imaging is also described in this paper.

1. Introduction

The state of the art echocardiographic modes for defining the endocardium of the left ventricle of the human heart are: fundamental imaging, second-harmonic imaging and left ventricle opacification. In common for these three modes is that the ultrasound probe contains an array of ferroelectric elements that allows the sound waves to be focused in defined directions by controlling the delays of emissions on each of these elements. Moreover, the probe is alternating between emitting focused sound pulses and receiving echos from different depths. Signals from different depths are separated by the different time of arrivals after the last pulse has been emitted. For each emitted beam, it is possible to receive more than one beam. The received beams are more focused and located within the emitted beam. A two-dimensional image is created by shooting beams in different directions.

The above mentioned imaging methods are all amplitude imaging methods because blood and tissue are separated by their difference in power [1]. When a signal is received at the probe it contains frequencies in the fundamental range and frequencies in the second-harmonic range. The fundamental frequency range is basically the same as the frequency content of the emitted pulse and the second harmonic frequency range is centered around two times the mean fundamental frequency. In second-harmonic imaging the strong signal with the fundamental freqencies are filtered out, while in fundamental imaging no filter is applied. In left ventricle

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opacification, blood is injected with a contrast agent and the amplitude of the second-harmonic frequencies are visualized.

Alternatively to the above mentioned methods, Doppler-based imaging methods can also be used to distinguish blood from tissue. By shooting many consecutive shots in the same direction, a vector of samples at each range cell is obtained. This is called the Doppler signal and is used to describe the movement inside each range cell. A Doppler signal from blood flow is different to a Doppler signal from tissue motion, since it is less coherent with depth. In Power Doppler, the blood is distinguished from the tissue signal by the power of the highpass filtered Doppler signal. Here, a packetsize of above 6 is required to achieve the desired filter characteristics [2]. The packetsize is the number of shots emitted in each direction and keeping this number low gives a better resolution.

In [3] bandwidth imaging has shown promising results in extracting the endocardial border in the left ventricle with a packetsize as small as 3. The small packetsize enables the method to have a resolution that is attractive for endocardial border definition.

Bandwith imaging distinguishes blood and tissue signals by the difference in coherence with depth between consecutive pulses and not by the difference in power. This method is not very sensitive to clutter noise in the nearfield of the probe and non-myocardial tissue is more brightly colored than in second-harmonic imaging. A drawback of the method is that tissue inside the ventricle such as trabeculae network, papillary muscles and valves are less visible. However, it was suggested in [3] that these effects might be advantageous in extracting the endocardial border.

In this paper, we seek to discuss the theoretical potential of knowledge-based imaging where the pulse strategy of the optimized bandwidth imaging method is taken as a starting point. The ultrasound recordings presented in [3] are re-used in this paper.

The definition of knowledge-based imaging relies on parametric models for the autocorrelation functions for signal from blood and tissue. In [4] a parametric model for the autocorrelation function for signal from laminar blood flow is outlined. In this paper, the assumption of laminar blood flow in the left ventricle is challenged.

Knowledge of the flow patterns in the left ventricle is documented in various ways. Three-dimensional particle traces of the left ventricle flow using time-resolved three-dimensional phase contrast MRI are shown in [5,6]. A computational flow dynamic model, where the initial boundary conditions are obtained by MRI is demonstrated in [7]. Moreover, in [8] a laboratory model of the left ventricle with prosthetic valves is investigated.

In this paper we are interested in the level of velocity spread we can expect inside one range cell in echocardiography. In [8] the authors argued that the Reynolds number, which is the ratio of inertial to viscous forces, is ranging from 7000 to 15,100 in the left ventricle. This means that the flow should be considered turbulent in the left ventricle.

In [9] a cascade model is introduced that explains how energy is transported from large eddies to smaller eddies where it is dissipated. Kolmogorov's microscales [10] are the smallest scales in turbulent flow. The length scale l_d is defined as

$$l_d = \left(\frac{\nu^3}{\epsilon}\right)^{1/4}, \tag{1}$$

where ν is the viscosity and ϵ is the average rate of dissipative energy per unit mass. If L is the turbulent length scale and U is the turbulent velocity scale of the largest eddies then the Reynolds number Re is equal to UL/ν. Since the kinematic energy is of the order U^2 and the turbulent time scale is of the order L/U, then ϵ is of the order U^3/L. We can therefore argue that:

$$\frac{l_d}{L} = \mathrm{Re}^{-(3/4)}. \tag{2}$$

If we choose L to be half of the ventricle long axis and we assume Re = 7000, then the minimum size of the eddies is at an order of 1/1500 of the ventricle long axis. This is at least 5 times smaller than the radial length of the range cell in echocardiography. Since the smallest eddies in the left ventricle flow can be completely inside one range cell, we postulate that the velocity spread inside one range cell can be substantial. We have therefore expanded the parametric model for the autocorrelation function in [4] to be valid for turbulent flow as well in Section 2.2.

In Section 3, general descriptions of the imaging methods are given and in Section 4, some premature knowledge-based images with different parameter settings are compared with a fundamental image, a bandwidth image and a second-harmonic image. In Section 5, knowledge-based imaging is defined and the apparent error rates of knowledge-based imaging, bandwidth imaging and fundamental imaging are calculated from computer generated signals in various types of noise.

For the record, [3] is an extended version of [11] and these two papers introduces bandwidth imaging. This paper introduces knowledge-based imaging and is an extended version of [12]. This paper is different than [12], since it contains a computer simulation, a more detailed theoretical derivation and a more extensive discussion of future work.

2. Signal model and data simulation

In [4] a parametric model for the autocorrelation functions in regions with laminar rectilinear flow is introduced. This model is valid under the assumption that the signal is a complex Gaussian process. In this section, this model is expanded to be valid for turbulent flow as well. The signal model is also further expanded to include additive white noise and clutter noise in a similar manner as in [13]. This is the theoretical framework for defining, instrumenting and discussing knowledge-based imaging and bandwidth imaging.

2.1. *Parametric model for the autocorrelation function of signal from blood and tissue*

As mentioned above the signal model of [4] is used. Here the authors assume a random continuum model for blood scattering [14]. The spatial fluctuation in mass density and