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Improvement of Automated Identification of the Heart wall by Reducing Stationary Clutter in Ultrasonic Echoes

静止クラッタ成分の低減による心臓壁領域の自動同定の性能向上

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1. Introduction

Currently, both cardiac wall motion and strain rate can be quantatively evaluated by means of ultrasound Doppler shift or speckle motion. In most of these methods, the heart wall, which is the object to be analyzed, is manually identified by an operator. However, this task is very time-consuming and suffers from inter- and intraobserver variability. With the aims of facilitation of analysis and elimination of operator dependence, automated identification of the heart wall needs to be realized. We have developed a method for automated identification of the heart wall region throughout an entire cardiac cycle by tracking the points classified in an initial frame. In our method [1], the multiple features, such as echogenicity and temporal phase changes of echo signals, are extracted to accurately identify heart wall in the initial frame.

RF echo signals in echocardiography contain the undesirable stationary clutters which are the components of echoes from external tissue such as the ribs [2]. In our previous study [1], to reduce this stationary clutter, high-pass filtering was applied to RF signals before extracting features, corresponding to the moving target indicator (MTI) filtering [3]. However, the magnitude-squared coherence (MSC) function of echoes, which is one of the features, still contained the effect of the stationary clutter. In this study, we improved the performance of the MSC by reducing the influence of the stationary clutter.

2. Principle

The MSC, which is obtained using complex spectra of RF echo signals in the tracked areas, evaluates the temporal variance in phase changes of echoes between two consecutive frames. In our study [1], the MSC at a frequency of 4.4 MHz, which was higher than the transmit frequency of 3.75 MHz, was used. Let us define the MSC as $|\gamma(i,j)|^2$ where i and j define the sampled point number corresponding to depth and the scan line number on an echocardiogram.

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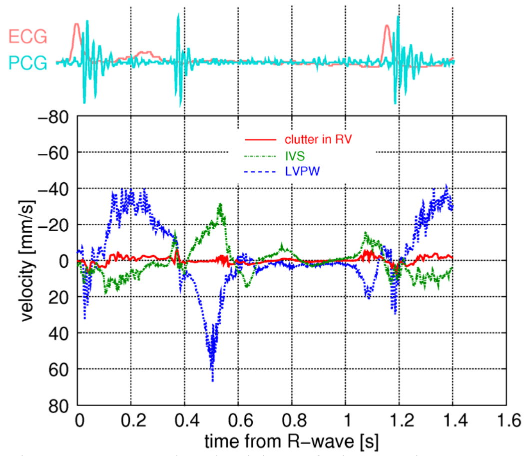

Fig. 1 Measured velocities of clutters in RV, IVS, and LVPW in echocardiography.

In general, the velocities of cardiac muscle and blood flow are much higher than that of the other organs, such as ribs, causing the clutter and, therefore, the clutter shows very small change in phase throughout a cardiac cycle [2]. **Figure 1** shows an example of the axial velocities of clutter in right ventricle (RV), interventricular septum (IVS) and left ventricular posterior wall (LVPW). In Fig. 1, the velocities were estimated by the *phased tracking method* [4] using echo signals, which were acquired at a frame rate of 751 Hz and contained the strong clutter caused by ribs. As shown in Fig. 1, the motion of clutter was much slower than that of the cardiac muscle. The information of the velocities of tissues is useful for reduction of influence of the stationary clutter.

The magnitude of velocity, $m(i, j)$, which was obtained by averaging velocities during a cardiac cycle, was estimated from echo signal by the *phased tracking method* before MTI filtering. A new MSC $G(i, j)$ was defined as the conventional MSC $|\gamma(i, j)|^2$ multiplied with clutter detection function $cdf(i, j)$ as follows:

$$G(i,j) = \frac{a}{1+c\exp\{-bm(i,j)\}}|\gamma(i,j)|^2$$
$$= cdf(i,j)|\gamma(i,j)|^2, \qquad (1)$$

where a, b, and c are the parameters of the logistic function. The value of $cdf(i, j)$, when $m(i, j) = 0$, is

$a / (1 + c)$. The inflection point occurs when $m(i, j) = \ln(a) / b$ and the value of $cdf(i, j)$ at this point is $a / 2$. In this study, a and $(1 + c)$ were set at 1.0 and 100.0, respectively, to remain the range of $|\gamma(i, j)|^2$ and to sufficiently decrease $G(i, j)$ when $m(i, j)$ is very low (the clutter would have substantial effect at this point (i, j)). Let us define the inflection point, $\ln(a) / b$, by the smallest average velocity $m_{\min}$ in those of the points inside the heart wall region. The points were selected based on the amplitude of MTI filtered echo signal. The echo amplitude of selected point was not less than the expectation amplitude e_1 of Rayleign distribution and not more than the amplitude e_2 when the cumulative probability of the distribution is 95%. The values of e_1 and e_2 were calculated from the MTI filtered echo signals in the area manually assigned as the heart wall region.

3. Comparison of results of identification using proposed and conventional MSC

The heart of a 23-year-old male was measured in longitudinal axis view at a very high frame rate of 860 Hz using parallel beamforming [5]. **Figures 2**(a) and 2(b) show B-mode image, which was constructed by detecting envelopes in beamformed RF signals sampled at 15 MHz, and the areas were manually assigned for determination of the initial parameters of expectation-maximization (EM) algorithm [6]. In addition, echo signals (before MTI filtering) in this area, which was assigned in the heart wall as shown in Fig. 2(b), were used for determination of the inflection point of $cdf(i, j)$. **Figures 3**(a) and 3(b) show the histogram of echo amplitude in the area assigned in the heart wall region shown in Fig. 2(b) and the extracted points, which follow the above condition, on the B-mode image. As shown in Fig. 3(b), the extracted points, which determine the inflection point of $cdf(i, j)$, were discretely distributed in the heart wall regions and not extracted from the lumen region.

Figures 4(a) and 4(b) show the conventional MSC and region-identified images. Figures 4(c) and 4(d) show the new MSC and region-identified images. Comparing Figs. 4(a) and 4(b) with Figs. 4(c) and 4(d), the increments of MSC of points in RV, which were caused by the clutter, were suppressed and correctly classified as lumen using the new MSC.

4. Conclusion

In this study, we proposed the modification of the MSC for reduction of influence of the stationary clutter. The influence of the stationary clutter was reduced using the MSC multiplied with the logistic function composed of proposed coefficients. In our previously proposed method [1], the heart wall region throughout the entire cardiac cycle would be identified more accurately using the new MSC using our proposed method.

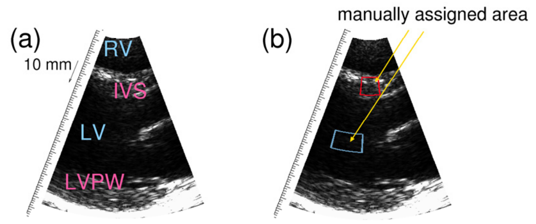

Fig. 2 (a) B-mode image and (b) areas manually assigned in heart wall region and lumen.

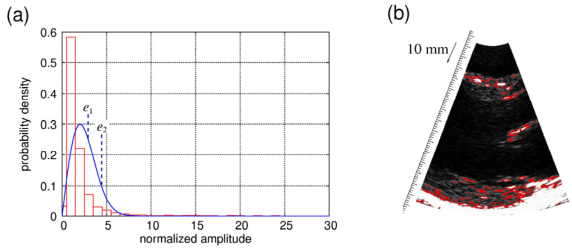

Fig. 3 (a) Histograms of echo amplitude in the area assigned in the heart wall region and the estimated Rayleigh distribution. (b) The extracted points (red points) for determining the inflection point of $cdf(i, j)$ overlaid on B-mode image.

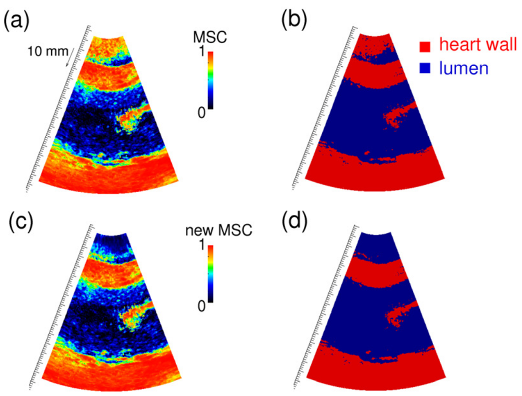

Fig. 4 (a) Conventional MSC image and (b) region-identified image using the conventional MSC. (c) New MSC image and (d) region-identified image using the new MSC.

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