Automated Segmentation of Heart Wall Based on Coherence Among Ultrasonic RF Echoes

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1. Introduction

In the conventional ultrasonic tomographic images, the echogenicity inside the heart wall is as low as that in the lumen. Moreover, there are many artifacts due to the reflection of side lobes by the external tissue such as ribs. Therefore, the heart wall cannot be distinguished from the cardiac lumen automatically using only the echogenicity. On the other hand, frequency characteristics of RF echoes contain more information (amplitude and phase at each frequency) on echoes from scatterers in comparison with the echo amplitudes as long as S/N is not too small. Therefore, there is possibility that the lumen and heart wall, whose echogenicities are similar are differentiated using frequency characteristics of RF echoes from them. In this study, complex frequency spectra of echoes from the heart wall and lumen were analyzed for the automated identification of those regions.

2. Methods

As shown in Fig. 1 (a), the interventricular septum (IVS), right ventricle (RV), and left ventricle (LV) were scanned by an ultrasonic beam, which scans sparsely to realize a high temporal resolution in a wide area, in the longitudinal-axis and apical 4-chamber views. RF data \(\{y_n(x)\}\) along each beam were acquired using a 3.75 MHz sector-type probe of ultrasonic diagnostic equipment (ALOKA SSD-6500), where \(n\) and \(x\) are frame (time) and depth, respectively. The sampling frequency of the RF signal was 15 MHz. The frame rate was 59 Hz in the longitudinal-axis scan and 546 Hz in the apical scan. RF data \(\{y_n(x)\}\) contain the components of echoes from the external tissue such as ribs due to side lobes which result in stationary artifacts. The artifact component was reduced by removing the direct current component from \(\{y_n(x)\}\) using the feature that the external tissue does not move so much with time.

Complex frequency spectrum \(Y_{n,x}(f)\) in each frame \(n\) and at each depth \(x\) was estimated by applying the short-time Fourier transform to the RF signal \(y_n(x)\) using a Hanning window (window length: 1.64 mm). Then, the temporal changes of the spectra were analyzed by applying the short-time Fourier transform to multiple frames. The window position was moved so as to follow the motion of a point of interest using the phased tracking method [1] in order to realize the analysis for the same scatterers during 100 frames and the spectra \(\{Y_{n,x}(f)\}\) in 100 frames were obtained for each position \(x\).

In the heart wall, the phase changes of RF echoes during a few millisecond are very small when the motion of the myocardium is tracked (compensated) because the myocardium stays in the focal area of the ultrasonic beam, whereas the small phase change would happen due to the change in thickness of myocardia. On the other hand, in the lumen, the motion of scatterers cannot be compensated because blood cells slip off from the focal area of the ultrasonic beam by blood flow. Therefore, phases of echoes would change even during a few milliseconds. In this study, the magnitude-squared coherence function (MSCF) \(\gamma_x(f)^2\) was employed for evaluating this difference in variance in phase changes of RF echoes. The MSCF \(\gamma_x(f)^2\) was expressed as follows:

\[
\gamma_x(f)^2 = \frac{E_m \left[ \tilde{Y}_{2n,x}(f) \tilde{Y}_{2n+N,x}(f) \right]^2}{E_m \left[ \tilde{Y}_{2n,x}(f) \right]^2 E_n \left[ \tilde{Y}_{2n+N,x}(f) \right]^2}, \quad (N = 1, 2, \ldots)
\]

where \(E_m[\cdot]\) and \(\ast\) show time (frame) averaging during 100 frames and complex conjugate, respectively.

For the automated identification of the heart wall using the MSCF \(\gamma_x(f)^2\), the optimal threshold level \(T_0(f)\) for the MSCF was determined based on the Bayes decision rule as follows:

\[
p(\gamma_x(f) | \Omega_1) P(\Omega_1) \leq p(\gamma_x(f) | \Omega_2) P(\Omega_2) \rightarrow |\gamma_x(f)|^2 \in \begin{cases} \Omega_1 \quad \Omega_2 \end{cases},
\]

where \(\Omega_1\) and \(\Omega_2\) are the myocardium and the lumen, respectively.

Fig. 1: (a) Illustration and (b) B-mode image of the human heart in its (1) longitudinal-axis view and (2) apical 4-chamber view. The regions surrounded by the red lines show the acquisition areas of RF signals.

Fig. 2: Histograms of brightness in the heart wall and lumen acquired from (a) 9 beams in the longitudinal-axis scan and (b) 10 beams in the apical 4-chamber scan.
where $\Omega_1$, $\Omega_2$, $P(\Omega_i)$, and $p(|\gamma_x(f)|^2|\Omega_i)$ show the class of heart wall, class of the lumen, a priori probability showing the occurrence of class $\Omega_i$, and the conditional probability density function that occurrence of $|\gamma_x(f)|^2$ belonging to class $\Omega_i$, respectively. In this study, the conditional probability density function $p(|\gamma_x(f)|^2|\Omega_i)$ and a priori probability $P(\Omega_i)$ were calculated based on the histograms of $|\gamma_x(f)|^2$ of the heart wall and the lumen which were manually segmented.

3. In vivo Experimental Results

RF data were acquired from the heart of a healthy 25-year-old male, and the analysis described in the previous section was applied. Central time in the analyzing period was set so that it corresponds to the rapid filling phase.

Figures 1(1-b) and 1(2-b) show B-mode images in the longitudinal-axis and apical views, respectively. The RF data were obtained for the beams shown by the arrows within the acquisition area surrounded by the red line in Fig. 1. Figures 2(a) and 2(b) show the histograms of normalized brightness in the longitudinal-axis and apical scans in the heart wall and lumen acquired from the beams in Fig. 1, respectively. The overlap of brightness acquired from the longitudinal-axis scan and from the apical scan between the heart wall and lumen was 40.2% and 43.8%, respectively. These large overlaps show that region identification of the heart wall using the brightness is difficult.

Figure 3 shows the MSCF $|\gamma_x(f)|^2$ at each depth $x$ along beam 5 in the data acquired from apical scan. In Fig. 3, the frame interval $N$ between two frames for the evaluation of the MSCF is 1 (1.69 ms) and 2 (3.38 ms), respectively. The MSCF of the RF echo scattered from the IVS was high. In contrast, the MSCF in the lumen was low.

Figure 4(a) and 4(b) show the examples of histograms of the MSCF $|\gamma_x(f)|^2$ in the heart wall and lumen acquired from 9 beams in the longitudinal-axis scan and from 10 beams in the apical scan, respectively. Figure 4 (c) show the overlap (%overlap) between the MSCF $|\gamma_x(f)|^2$ of the heart wall and the lumen. These histograms correspond to probability density functions $p(|\gamma_x(f)|^2|\Omega_i)$ ($i=1, 2$) and the %overlap shows the probability of misclassification in eq. (2). The possibility of misclassification is the smallest (3.0% and 13.3% in the data acquired from the longitudinal-axis and apical scan, respectively) when $N = 1$ and 4.2 MHz. Based on eq. (2), the optimal threshold $T_0(f)$ for the MSCF was determined as shown in Fig. 4(d). Subsequently, the results of the region identification are shown in Fig. 5(c). In the longitudinal view, the IVS is clearly differentiated from the RV and the LV in all cardiac phases and aortic valve (AV) can be also observed as shown in Fig. 5 (1-c). Although, in the apical view, the interfaces of the IVS are not clear due to the low resolution along lateral direction, the almost all parts of the IVS are differentiated from the RV and the LV as shown in Fig. 5 (2-c).

4. Conclusions

The possibility of the automated region identification of the heart wall using the magnitude-squared coherence function was shown by the in-vivo experimental results that the heart wall can be clearly differentiated from the lumen when the optimal threshold for the magnitude-squared coherence function is determined.

References