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Myocardial Strain Imaging System with a High-Performance Adaptive Dynamic Grid Interpolation Method
高性能の適応的動的格子補間法を使用する心筋歪みイメージングシステム

Shuhui Bu[1 †], Makoto Yamakawa[2], and Tsuyoshi Shiina[1] ([1]Faculty of Med., Kyoto Univ.; [2]Adv. Bio. Eng. Res. Unit, Kyoto Univ.)
布 樹輝[1†], 山川 誠[2], 椎名 毅[1]
([1]京都大 医学科; [3]京都大 先端医工学)

1. Background

Decreases of myocardial motion often appear in the early stage of ischemic heart disease. For the complex 3-D myocardial motion, 3-D assessment of the myocardial contractile performance is required for accurate diagnosis. We previously proposed a new method of 3-D myocardial motion tracking and visualization of the invariant of a full strain tensor using a 2-D phased array. The feasibilities of the proposed methods were evaluated by numerical simulations[1]. However, the calculated spatial gradient of displacement (strain) is fairly sensitive to noise. Therefore, smoothing filters are applied before strain calculation. Traditional smoothing filters suffer from a critical drawback; there is a trade-off between spatial resolution and stability. To overcome the problem, we proposed a novel displacement smoothing filter which called adaptive dynamic grid interpolation (ADGI) method for reducing the effects of noise in strain calculation. The performance of propsed method is evaluated and finite element method is used to simulate the strain distribution in the myocardium. In a model with the infarcted region located around 1 to 3 o'clock, the RMS error is improved from 41.47% to 22.90% without degrading spatial resolution.

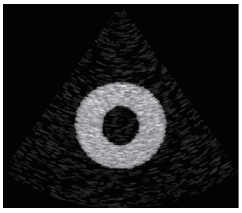
2. Method

In this system, a 2-D transducer is used for acquiring echo data from the myocardium by volumetric sector scanning. Radio frequency (RF) signals for each scan line are received at all elements in the probe. The phase-shifts at every measuring point between several consecutive frames are calculated by the combined autocorrelation method (CAM) or the extended CAM[3], and then the displacement vectors are calculated by weighted phase gradient method[1,4] from obtained phase-shifts. At the same time, the B-mode images, which are generated by the RF signals obtained from the aperture in the center transducer, are acquired. From the B-mode images, the myocardial endocardium boundaries are segmented, while the epicardium boundary is calculated from the endocardium boundary by adding a fixed radius. Next, the myocardium meshes are generated from the myocardial boundaries. In the following step, the un-smoothed displacement vectors and myocardial meshes are input into a displacement re-sampling module. The displacement vectors in each mesh's node are calculated by bilinear interpolation, and their coordinates are also translated from Cartesian coordinates to cylinder coordinates. After this, the ADGI method is used for smoothing the displacement vectors. Finally, the strain tensors are calculated from the smoothed displacement vectors.

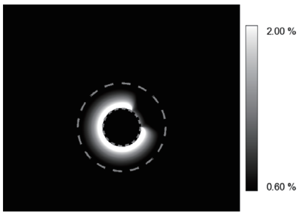
In the ADGI method, the displacement vectors in all sampling nodes are defined as $\vec{u} = (u_r, u_c)$, where u_r is radial direction's displacement component and u_c is circumference direction's displacement component. Virtual springs are attached at each mesh node in radial and circumference directions, respectively. Through the virtual springs, revising displacements $\vec{\varepsilon} = (\varepsilon_r, \varepsilon_c)$ are introduced, and an overall error function is defined as:

$$
\begin{aligned}
e(\varepsilon_r, \varepsilon_c) &= \sum_{i=1}^{n}\sum_{j=1}^{m}\left[(\varepsilon_r^{\,i,j})^2 + (\varepsilon_c^{\,i,j})^2\right] \\
&+ \sum_{i=1}^{n}\sum_{j=1}^{m-1}\frac{E_{rr}^{i,j}}{2}(u'^{\,i,j+1}_r - u'^{\,i,j}_r)^2 \\
&+ \sum_{i=1}^{n-1}\sum_{j=1}^{m-1}\frac{G_{rc}^{i,j}}{2}\left[(u'^{\,i+1,j+1}_r - u'^{\,i+1,j}_r) - (u'^{\,i,j+1}_r - u'^{\,i,j}_r)\right]^2 \\
&+ \sum_{i=1}^{n-1}\sum_{j=1}^{m}\frac{E_{cc}^{i,j}}{2}(u'^{\,i+1,j}_c - u'^{\,i,j}_c)^2 \\
&+ \sum_{i=1}^{n-1}\sum_{j=1}^{m-1}\frac{G_{cr}^{i,j}}{2}\left[(u'^{\,i+1,j+1}_c - u'^{\,i,j+1}_c) - (u'^{\,i+1,j}_c - u'^{\,i,j}_c)\right]^2,
\end{aligned}
$$

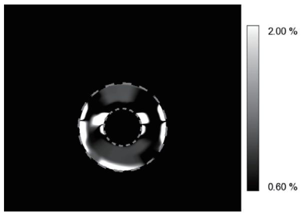
where m is the radial direction's node number, n is the circumference direction's node number, and



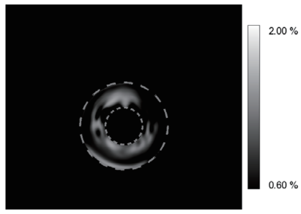
(a) B-Mode image



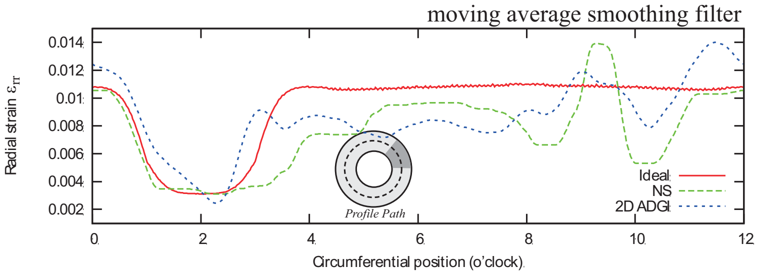
(b) Ideal radial strain



(c) Radial strain generated by moving average smoothing filter



(d) Radial strain generated by ADGI



(e) Profiles of radial strain generated from two consecutive frames

Fig. 1 Resulting images and profiles

$u' = (u'_r, u'_c)$ is the revised displacement vectors.

By minimizing the error function, the revised displacements, which combine original and revising displacements at each mesh node, are obtained. The virtual spring's pseudo-elasticity parameters E_{rr} and E_{cc} control the axial displacement revision. The pseudo shear elasticity parameters G_{rc} and G_{cr} control the shear direction's revising effect. The pseudo-elasticity parameters are calculated from displacement error functions, which indicate the magnitude of error motion. The displacement error functions are defined as:

$$\delta_{rr}^{i,j} = \left| \frac{u_r^{i,j+1} - 2u_r^{i,j} + u_r^{i,j-1}}{h_r^{i,j} + h_r^{i,j+1}} \right|,$$

$$\delta_{cc}^{i,j} = \left| \frac{u_r^{i+1,j} - 2u_r^{i,j} + u_r^{i-1,j}}{h_r^{i,j} + h_r^{i+1,j}} \right|,$$

$$\delta_{rc}^{i,j} = \delta_{cr}^{i,j} = \frac{\delta_{rr}^{i,j} + \delta_{cc}^{i,j}}{2},$$

where h_r and h_c are the distance between adjacent nodes in the radial and circumferential directions, respectively. The pseudo-elasticity parameters are mapped to the range [E^{min} E^{max}] by a *sin* function. Finally, the strain tensors are calculated from the smoothed displacement vectors.

3. Simulation

The performance of our proposed method is evaluated by numerically simulating the short-axis imaging of a 3-D myocardial model. The strain distribution is simulated by using finite element method. The hypothetic infarcted wall is located around 1 to 3 o'clock. Young's modulus in a normal wall is taken as 75 kPa and 225 kPa in an infarcted wall, and Poisson's ratio is set to 0.48. Pressure applied to endocardium is 11 mmHg. The 2-D transducer's outer diameter is 20 mm. The ultrasonic pulse has a center frequency of 3.75 MHz, and the fractional bandwidth is 40%. The signal-to-noise ratio is set to 20 dB. The optimized pseudo-elasticity parameters in the radial direction are set in the range of 20 to 2800 in the radial direction and 20 to 800 in the shear direction, respectively.

4. Results and Conclusions

The resulting images and profiles are shown in Figure 1. Figure 1(a) shows the B-mode image. Figure 1(b) shows the ideal radial strain, figure 1(c) shows the radial strain generated by applying moving average filter, and figure 1(d) shows the radial strain generated by applying ADGI method. Figure 1(e) shows the profiles of radial strain along the mid-wall. The RMS error of images generated from moving average filter is 41.47%, and the RMS error of image (d), which is generated by our proposed method, is 22.90%. From the resulting images and evaluating parameters, we can see that even the SNR is low, our proposed method still can output accurate and stable myocardial strain image.

References

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