Polar-Regularized Left Ventricular Strain Analysis from Cine MRI Using Non-rigid Registration

W. Feng[1], H. Gupta[2], S. G. Lloyd[2], L. J. Dell'Italia[2], and T. S. Denney Jr. [1]

[1]Electrical and Computer Engineering, Auburn University, Auburn University, AL, United States, [2]University of Alabama at Birmingham

INTRODUCTION

Quantitative measurements of left ventricular (LV) strain are important in the diagnosis and management of patients with heart disease and tracking the efficacy of treatments over time. While tagged and DENSE cardiac magnetic resonance imaging (MRI) are established methods for measuring LV strain, cine MRI is the most commonly-used cardiac MRI protocol, where user-defined contours are drawn at end-diastole (ED) and end-systole (ES) to compute volumes and ejection fraction. These contours contain information about myocardium deformation that should be used in the strain estimate. In [1], a new method for computing two-dimensional (2D) LV strain from cine cardiac MRI called cine myocardial deformation analysis (CMDA) is proposed. The CMDA method is based on non-rigid registration and incorporates information from user-defined contours. The CMDA algorithm in [1], however, used a Cartesian coordinate deformation model with no smoothness constraints. As a result, the inherent smoothness in a B-spline model imposed equal smoothness in the two Cartesian coordinate directions, which do not always correspond to the primary directions of cardiac wall deformation. In this abstract, an improved version of the CMDA algorithm is presented that applies a polar-coordinate smoothness constraint. With polar coordinate regularization, different amounts of smoothing can be applied in the radial and circumferential directions, which are the primary directions of cardiac deformation.

THEORY

In CMDA, myocardial deformation is estimated in a two-stage process. In the first stage, non-rigid registration is used to compute inter-frame deformation fields between timeframes [2]. The inter-frame deformation fields are then used to propagate the ED and ES contours to all other timeframes. In the second stage, the inter-frame deformation fields are refined by re-registering the images with a regularization term based on the propagated contours [1]. Finally strains are computed from the refined motion estimate.

In both rounds of inter-frame registration, the inter-frame deformation is computed by optimizing the cost function $E = E_{sim} + E_{creg} + E_{polar}$, where E_{sim} and E_{creg} are image data consistency term and contour regularization terms defined in [1] and E_{polar} is a polar-coordinate regularization term defined as follows. Let $\mathbf{m}(\mathbf{x})$ be the Cartesian deformation. Let $\mathbf{r}(\mathbf{x})$ and $\mathbf{n}(\mathbf{x})$ be the radial and circumferential unit vectors at $\mathbf{x}$. Then the projections $m_r = \langle \mathbf{m}(\mathbf{x}), \mathbf{r}(\mathbf{x}) \rangle$ and $m_c = \langle \mathbf{m}(\mathbf{x}), \mathbf{n}(\mathbf{x}) \rangle$ are the deformations in radial and circumferential directions. The polar regularization term is $E_{polar} = \int \alpha \left\| \Delta m_r(\mathbf{x}) \right\|^2 + \beta \left\| \Delta m_c(\mathbf{x}) \right\|^2 d\mathbf{x}$, where $\Delta m_r(\mathbf{x})$ and $\Delta m_c(\mathbf{x})$ are the Laplacians of $m_r(\mathbf{x})$ and $m_c(\mathbf{x})$ in radial and circumferential directions. α and β are weighting parameters that control the amount of radial and circumferential regularization. α and β where empirically chosen to be 10^{-7} and 10^{-6} respectively, which reflects an assumption that the LV typically thickens more in the radial direction than it contracts in the circumferential direction.

METHODS

38 normal human volunteers (NL) and 42 patients with myocardial infarction (MI) were imaged with an SSFP sequence with the following parameters: TR 3.8, TE 1.6 ms, slice thickness 8mm, no inter-slice spacing, flip angle 45 deg, k-space segmentation 10 views per segment, matrix 256x128, field of view 42 cm, 1 signal average, bandwidth 125kHz. LV contours were drawn at ED and ES for all valid short axis slices semi-automatically by trained experts. CMDA radial and circumferential strains were then computed in all studies. A training set of 18 NL and 22 MI patients was used to choose α and β. An evaluation set of 20 NLs and 20 MI were used to compare CMDA circumferential strains with strains computed using HARmonic Phase (HARP) analysis [3] and a 3D model-based method [4]. Radial CMDA strains were not compared with HARP or 3D model-based analysis because radial strains computed from tagged MRI were unreliable.

RESULTS AND DISCUSSION

Fig 1 shows plots of circumferential strain versus time computed using CMDA and HARP averaged over the 20 normal human volunteers in the evaluation set. Time was normalized in these plots so that zero corresponds to end-diastole and 100 corresponds to early diastole. CMDA agrees with HARP at all levels, but the agreement is particularly good at mid-ventricle. Strains past early diastole are not shown because HARP strains were not reliable past early diastole due to tag fading.

Fig. 2 shows CMDA strain overlaid on a mid-ventricular slice for all imaged time frames in a normal human volunteer. As expected, contraction increases through systole and decrease during diastole. Strain is greater in the lateral wall than the septal wall, and strain is greater near the endocardium than near the epicardium

Table I shows correlation coefficients between mid-ventricular CMDA and HARP and 3D model-based strains computed over the 20 NLs and 20 MI subjects in the evaluation set. Only end-systolic strains were computed for the 3D model-based method, but they showed good correlation with peak CMDA strain.

CONCLUSION

While tagged and DENSE MRI will continue to be the gold standards for measuring LV strain, LV strain can be accurately measured from cine MR cardiac images using CMDA. Clinically, strain measured from cine MRI may be more useful than strain from tagged or DENSE MRI because cine MRI is more widely used in this environment. In addition, cine MRI typically has higher temporal resolution than tagged or DENSE MRI. Finally, CMDA strain has the potential to compute late diastolic strains that can be difficult to compute with tagged or DENSE MRI.

REFERENCES

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Table I: Correlation coefficients (ρ) between mid ventricular peak CMDA circumferential strain and HARP strain and 3D model-based strain.

	ρ	P
HARP Peak Strain	0.72	<0.001
HARP Peak Systolic Strain Rate	0.61	<0.001
HARP Peak Early Diastolic Strain Rate	0.65	<0.001
3D Model-Based Strain	0.69	<0.001

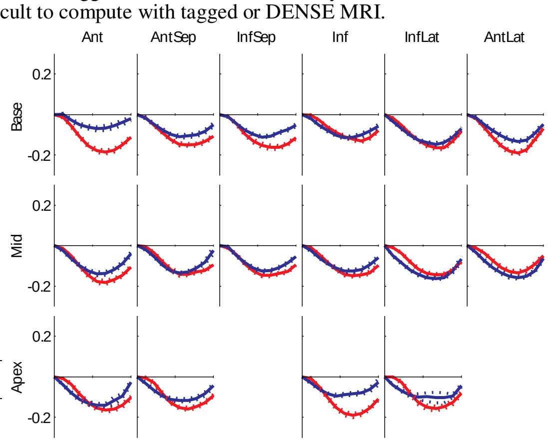

Fig. 1: Circumferential strains vs. time computed from cine MRI (red) and tagged MRI/HARP (blue). The curves represent the average (solid) and ± standard error (dotted) over 20 normal human volunteers.

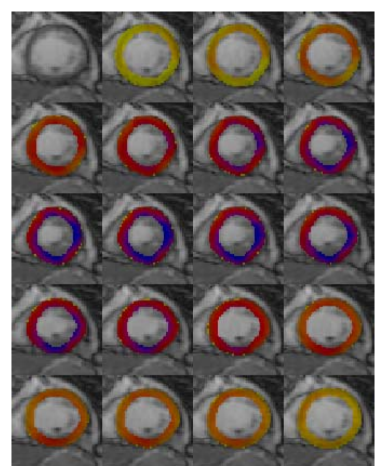

Fig. 2: Circumferential strain overlaid on a mid-ventricular slice of a normal human volunteer. Yellow represents no contraction, red represents 10% contraction and blue represents 20% contraction. Time increases from left to right starting from the upper-left corner.