Cardiac Motion Estimation by Optimizing Transmural Homogeneity of the Myofiber Strain and Its Validation with Multimodal Sequences

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Abstract. Quantitative motion analysis from cardiac imaging is important to study the function of heart. Most of existing image-based motion estimation methods model the myocardium as an isotropically elastic continuum. We propose a novel anisotropic regularization method which enforces the transmural homogeneity of the strain along myofiber. The myofiber orientation in the end-diastolic frame is obtained by registering it with a diffusion tensor atlas. Our method is formulated in a diffeomorphic registration framework, and tested on multimodal cardiac image sequences of two subjects using 3D echocardiography and cine and tagged MRI. Results show that the estimated transformations in our method are more smooth and more accurate than those in isotropic regularization.

Keywords: Cardiac motion estimation, diffeomorphic registration, cardiac strain, myofiber orientation.

1 Introduction

Cardiac motion estimation is important to study the heart function and its disease mechanism. Cardiac images such as 3D echocardiography, cine-MRI and cardiac CT have been widely used for quantitative motion analysis. However, due to the low spatial and temporal resolution and the complexity of the cardiac biomechanics, accurate cardiac motion estimation is still a challenging problem.

Cardiac motion estimation is generally solved by using nonrigid registration methods. To make the deformation unique, various regularization methods have been proposed to utilize the prior knowledge of the myocardium shape and dynamics. Spatiotemporal smoothness has been used to regularize the deformations [1–3]. Particle trajectory smoothness, transformation symmetry and transitivity have been used to constrain the temporal smoothness [4, 5]. Diffeomorphic image registration has been proposed to estimate transformation as the end of a smoothly evolving process constrained by a differential equation [6] and it has been extended to motion estimation from multiple frames [7, 8]. Anatomical constraint such as myocardium incompressibility has also been used in [8, 9].

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In reality the myocardium has a fibrous structure and the myocardium motion are closely related to the myofiber orientation [10, 11]. Little effort has been made to incorporate the myofiber orientation as a constraint for motion estimation. Papademetris *et al.* [12] registered the consecutive frames by using a regularization term consisting of a strain energy of the transversely isotropic myofiber model. In our previous work [13], an anisotropic regularization is proposed to make the velocity field more smooth in the myofiber direction.

In this paper, we propose a regularization term to enforce the transmural homogeneity of myofiber strain. It is deduced from the knowledge that the mechanical load is distributed uniformly in the myocardium of healthy subjects [14]. Experiments have shown that optimization of the myofiber strain homogeneity can lead to reasonable myofiber orientations [15]. Tseng *et al.* [16] has shown reversely that the myofiber strain is transmurally uniform. We implement the proposed method in a diffeomorphic registration framework in which the velocity field is regularized to make the myofiber strain transmurally uniform. The myofiber orientation is obtained by registering the reference frame of the sequence with a diffusion tensor (DTI) myofiber atlas. We validate the proposed method with echocardiography and cine-MRI images of two healthy subjects. Results show that our method can derive transformations with more smooth transmural strain and higher tracking accuracy than isotropic regularization methods.

2 Method

2.1 Diffeomorphic Motion Estimation

We adopt a diffeomorphic registration for our motion estimation method. The deformation between time zero to t is defined by a velocity field using a differential equation of $\frac{d\boldsymbol{\phi}}{dt} = \boldsymbol{v}(\boldsymbol{\phi}(\boldsymbol{x},t),t), \boldsymbol{\phi}(\boldsymbol{x},0) = \boldsymbol{x}$, with $\boldsymbol{x} \in \Omega \subset R^3, t \in [0,T]$ and $T = N_s - 1$ where N_s is the number of frames. The motion estimation problem is stated as an optimization of a variational energy of the velocity field $\boldsymbol{v}(\boldsymbol{x},t)$:

$$\hat{\boldsymbol{v}} = arg \inf_{\boldsymbol{v} \in V} \lambda \int_0^T E_{reg}(\boldsymbol{v}) dt + \sum_{n=1}^{n=T} E_{SSD}(I_{n-1}(\boldsymbol{x}), I_n(\boldsymbol{\phi}_{n-1,n})). \quad (1)$$

The first term is a regularizer to evaluate the spatiotemporal smoothness. The second term is a similarity measurement which evaluates the summed squared difference (SSD) between I_{n-1} and the unwarped frame $I_n(\boldsymbol{\phi}_{n-1,n})$, with $\boldsymbol{\phi}_{n-1,n}$ being the deformation between $(n-1)$th and nth frames. λ is the weighting to balance these two energy terms.

We parameterize the velocity field [8] and the solution of Eqn.(1) is obtained by numerical optimization. The velocity field is defined by a series of 3D B-spline functions at time $t_k (k = 0, 1, ..., N_t, t_k = k\Delta t, \Delta t = 1/N_f)$, with N_f being the number of time steps between two consecutive frames, $N_t = N_f \times T$ being the total number of B-spline functions. The velocity field function at time point t_k is defined as $\boldsymbol{v}(\boldsymbol{x}, t_k) = \sum \mathbf{c}_{m;k} \boldsymbol{\beta}(\boldsymbol{x} - \boldsymbol{x}_m)$, with $\mathbf{c}_{m;k}$ being the control vectors located on a uniform grid of $\boldsymbol{x}_m$ at t_k, $\boldsymbol{\beta}(\boldsymbol{x} - \boldsymbol{x}_m)$ being the 3D B-spline kernel

function at $\boldsymbol{x}_m$. The transformation $\boldsymbol{\phi}(\boldsymbol{x},t)$ is expressed as the forward Euler integral of the velocity field $\boldsymbol{v}(\boldsymbol{x},t)$ by assuming that the velocity of each point is piecewise constant within a time step. The diffeomorphism $\boldsymbol{\phi}_{n-1,n}$ is estimated by composition of all the small deformation defined by the velocity field between time $t_{(n-1)*N_f}$ and t_{n*N_f}:

$$\boldsymbol{\phi}_{n-1,n} = \prod_{(n-1)N_f}^{nN_f-1} (\mathbf{Id} + \boldsymbol{v}_k), \tag{2}$$

with $\mathbf{Id}$ the identity transformation and $\boldsymbol{v}_k = \boldsymbol{v}(\boldsymbol{x}, t_k)$ the velocity field at t_k. By using Eqn.(2), each SSD term in Eqn.(1) only depends on the control vectors within the time of consecutive frames. Thus the derivative of the SSD with respect to the control vectors can be evaluated independently.

2.2 Regularization of Myofiber Strain along Transmural Direction

The regularization energy consists of spatial and temporal terms. We denote the displacement field with $\boldsymbol{u}(\boldsymbol{x}) = (u_1(\boldsymbol{x}), u_2(\boldsymbol{x}), u_3(\boldsymbol{x}))^t$ and $\boldsymbol{x} = (x_1, x_2, x_3)^t$. Infinitesimal strain is used in our motion analysis because the deformation between two consecutive frames is usually small. The strain tensor $\boldsymbol{\epsilon}$ is defined as:

$$\boldsymbol{\epsilon} = \begin{pmatrix} \frac{\partial u_1}{\partial x_1} & \frac{1}{2}(\frac{\partial u_1}{\partial x_2} + \frac{\partial u_2}{\partial x_1}) & \frac{1}{2}(\frac{\partial u_1}{\partial x_3} + \frac{\partial u_3}{\partial x_1}) \\ \frac{1}{2}(\frac{\partial u_2}{\partial x_1} + \frac{\partial u_1}{\partial x_2}) & \frac{\partial u_2}{\partial x_2} & \frac{1}{2}(\frac{\partial u_2}{\partial x_3} + \frac{\partial u_3}{\partial x_2}) \\ \frac{1}{2}(\frac{\partial u_3}{\partial x_1} + \frac{\partial u_1}{\partial x_3}) & \frac{1}{2}(\frac{\partial u_3}{\partial x_2} + \frac{\partial u_2}{\partial x_3}) & \frac{\partial u_3}{\partial x_3} \end{pmatrix}, \tag{3}$$

with $\epsilon_{i,j} = \frac{1}{2}(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i})$ being the (i,j)th component of $\boldsymbol{\epsilon}$. Given the myofiber orientation at point $\boldsymbol{x}$ as $\boldsymbol{f} = (f_1, f_2, f_3)^t$, the myofiber strain at $\boldsymbol{x}$ is evaluated as:

$$\boldsymbol{\epsilon_f} = \boldsymbol{f}^t \boldsymbol{\epsilon} \boldsymbol{f} = \sum_{i,j} \epsilon_{i,j} f_i f_j. \tag{4}$$

We define a function which evaluates the myofiber strain variation along transmural direction $\boldsymbol{\tau} = (\tau_1, \tau_2, \tau_3)^t$ by using the directional derivative:

$$V_{\epsilon_{\boldsymbol{f}}, \boldsymbol{\tau}} = \frac{\partial \boldsymbol{\epsilon_f}}{\partial \boldsymbol{x}} \boldsymbol{\tau} = \sum_{i,j,p} \frac{\partial^2 u_p}{\partial x_i \partial x_j} f_i f_j \tau_p. \tag{5}$$

Eqn.(5) defines a general form of the strain variation along any direction. We can see that each of the isotropic bending energy term used in [1] evaluates the variation of normal and shear strains along x_1, x_2, x_3 directions. We define a regularization term which minimizes the anisotropic myofiber strain variation transmurally on the myocardium ($\boldsymbol{x} \in \Omega_M$), and the isotropic strain variations at points outside the myocardium ($\boldsymbol{x} \in \bar{\Omega}_M$) by:

$$E_{sr} = 27 \int_{\Omega_M} \sum_{i,j,p} (\frac{\partial^2 u_p}{\partial x_i \partial x_j} f_i f_j \tau_p)^2 d\boldsymbol{x} + \int_{\bar{\Omega}_M} \sum_{i,j,p} (\frac{\partial^2 u_p}{\partial x_i \partial x_j})^2 d\boldsymbol{x}, \tag{6}$$

where 27 is used to weight the anisotropic term since the isotropic term has 27 combinations of the strains and directions of derivative. Ω_M is domain of

myocardium points. We apply the spatial regularization to each velocity field $\boldsymbol{v}_k$ to maximize the homogeneity of myofiber strain along transmural direction in the whole cardiac cycle.

The temporal regularization term evaluates the magnitude of first order derivative of the point velocity with respect to time, which is defined as:

$$E_{tr} = \int_{\Omega} |\boldsymbol{v}_k(\boldsymbol{x} + \boldsymbol{v}_{k-1}\Delta t) - \boldsymbol{v}_{k-1}|^2 d\boldsymbol{x}. \tag{7}$$

The total regularization energy in Eqn.(1) is defined as a weighted sum of the spatial and temporal regularization energy at all time points:

$$E_R = \int_0^T E_{reg} dt = \sum_{k=0}^{N_t} E_{sr}(\boldsymbol{v}) + w_t \sum_{k=1}^{N_t} E_{tr}(\boldsymbol{v}). \tag{8}$$

2.3 Optimization

We use a steepest descent method to optimize the parameterized function. The derivative of $E_{SSD}(I_{n-1}(\boldsymbol{x}), I_n(\boldsymbol{\phi}_{n-1,n}))$ with respect to the control vectors $\mathbf{c}_{m;k}, ((n-1) * N_f \leq k < n * N_f)$ is:

$$\frac{\partial E_{SSD}}{\partial \mathbf{c}_{m;k}} = \int_{\Omega_s} (I_n(\boldsymbol{\phi}_{n-1,n}) - I_{n-1}) \nabla I_n(\boldsymbol{\phi}_{n-1,n}) \frac{\partial \boldsymbol{\phi}_{n-1,n}}{\partial \mathbf{c}_{m;k}} d\boldsymbol{x}, \tag{9}$$

with Ω_s being the domain which is controlled by $\boldsymbol{c}_{m;k}$ and for other value of k the gradient is zero. For the derivative of the spatial regularization with respect to the pth component of $\boldsymbol{c}_{m;k}$, we have:

$$\frac{\partial E_{sr}}{\partial c_{m,p;k}} = \int_{\Omega_s} \sum_{i,j} V^k_{\epsilon_{i,j,p}} \frac{\partial^2 \beta_p(\boldsymbol{x} - \boldsymbol{x}_m)}{\partial x_i \partial x_j} d\boldsymbol{x}, \tag{10}$$

with $V^k_{\epsilon_{i,j,p}}$ being one summand of myofiber strain variation along transmural direction in Eqn.(5) at time point k and $\beta_p(\cdot)$ being the pth component of the B-spline function. The derivative of the temporal regularization term is approximated by assuming small displacement between two time steps:

$$\frac{\partial E_{tr}}{\partial c_{m,p;k}} = \int_{\Omega_s} (2 * v_{m,p;k} - v_{m,p;k-1} - v_{m,p;k+1}) \beta_p(\boldsymbol{x} - \boldsymbol{x}_m) d\boldsymbol{x}. \tag{11}$$

2.4 Myofiber and Transmural Directions on Myocardium

We use a human diffusion tensor image (DTI) [1] as the myofiber atlas. The myofiber direction at each DTI voxel is defined as the eigenvector of the diffusion tensor corresponding to the largest eigenvalue. The myofiber orientation is then mapped into the reference frame by using a transformation estimated from a nonrigid registration with the anatomical image of the DTI. The transmural direction at each myocardium point is estimated from the myofiber orientation. It is defined as a vector in the short axis plane and is perpendicular to the myofiber. The myofiber and transmural directions in the reference frame are propagated to each of the following frames by using the recovered deformation.

[1] `http://www.ccbm.jhu.edu/research/dSets.php`

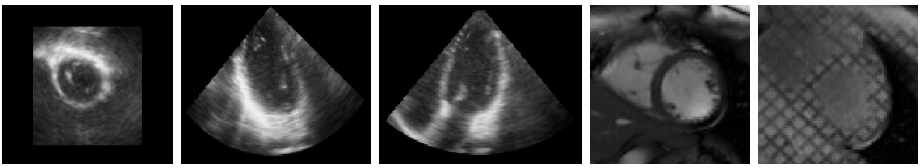

Fig. 1. The ED frame of echocardiography and MRIs of one subject. The first three images are orthogonal view of the echocardiography. The last two images are short axis ROIs from the cine MRI and the tagged MRI images.

3 Datasets and Experiments

We validated our method using multimodal image sequences of two healthy subjects. In the first experiment, an echocardiographic sequence of each subject was acquired with a Philips IE33 system. The number of frames is 26 and the frame size was $135 \times 126 \times 106$ with voxel size $1.12mm^3$. In the second experiment, both cine MRI and tagged MRI of each subject were acquired with a Siemens system. The number of frames in both sequences was 30 and the frame size was $256 \times 256 \times 10$ with voxel size $1.25mm \times 1.25mm \times 8mm$. Fig.1 shows the images of one subject. In both experiments, we compared the proposed method with a method using isotropic regularization [1]. In first experiment, we evaluated the strains in the 16 AHA segments [17] (apex segment was excluded). Both epicardial and endocardial surface meshes in the end-diastole (ED) frame were generated by automatically labeling the short axis contours and connecting the points into triangle meshes. Eight meshes of intermediate layers were interpolated from the endocardial and epicardial meshes. The radial, circumferential and longitudinal directions in each vertices were automatically calculated as did in [8] and the myofiber direction was estimated by using the transformation from the ED frame to the DTI atlas. The cross-fiber direction was defined as the cross product of the radial and myofiber directions. The myofiber, cross-fiber, circumferential and longitudinal strains were evaluated and their variations along the ten transmural layers in each segments are plotted and compared.

In the second experiments, the cine-MRI images were used to estimate the cardiac motion. The estimated transformation was used to deform the grid crossing points extracted from the ED frame of the tagged MRI. The affine transformation between the tagged MRI and the cine MRI was acquired from the DICOM file headers. We compared the two methods by overlaying the deformed ED tagging grid on the ES frames.

4 Results

We use a series of 3D B-spline functions with spatial spacing of 6 and temporal spacing of 1. The weightings λ and w_t for the regularization terms are set to 0.1 and 0.005 for both methods.

The results for both subjects are similar and we show the results of first subject as an example. We first show the strain estimation in the echocardiography sequence in Fig.2. By comparing the first two rows, we can see that the strains in our proposed method are more temporally smooth than the method with isotropic regularization. It indicates that the estimated myocardium motion is more uniform in our method. It is more consistent with the cardiac motion of a healthy subject. The errorbars in the bottom row show the average and transmural variance of the myofiber strains in four mid cavity segments. We can see that our estimated motion can lead to transmural homogeneity of the myofiber strain (in blue). Similar result can be seen in the cine MRI experiment (Fig.3). The result of the second experiment is shown in Fig.4. We overlay the deformed tagging grids in both apical and basal planes of ED frame on the end-systolic frame. We can see the improvement of our method over the isotropic regularization method.

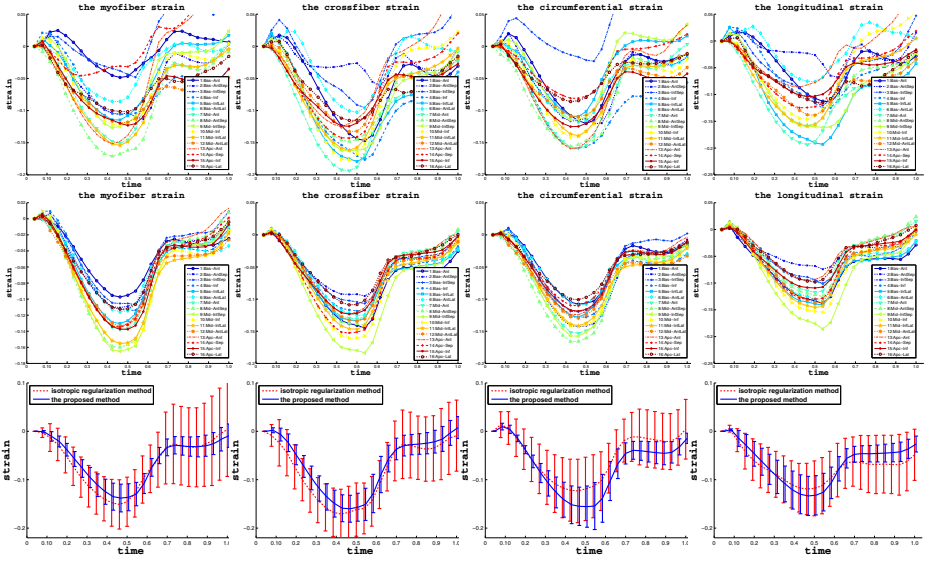

Fig. 2. The strains in the echocardiography sequence. (Top row) Strains in the isotropic regularization method. (Middle row) Strains in our method. (Bottom row) Average and transmural variance of the myofiber strains in four segments using isotropic regularization method (red) and our method (blue). The four segments are the anterior, anteriorseptal, inferior and anteriorlateral segments, respectively.

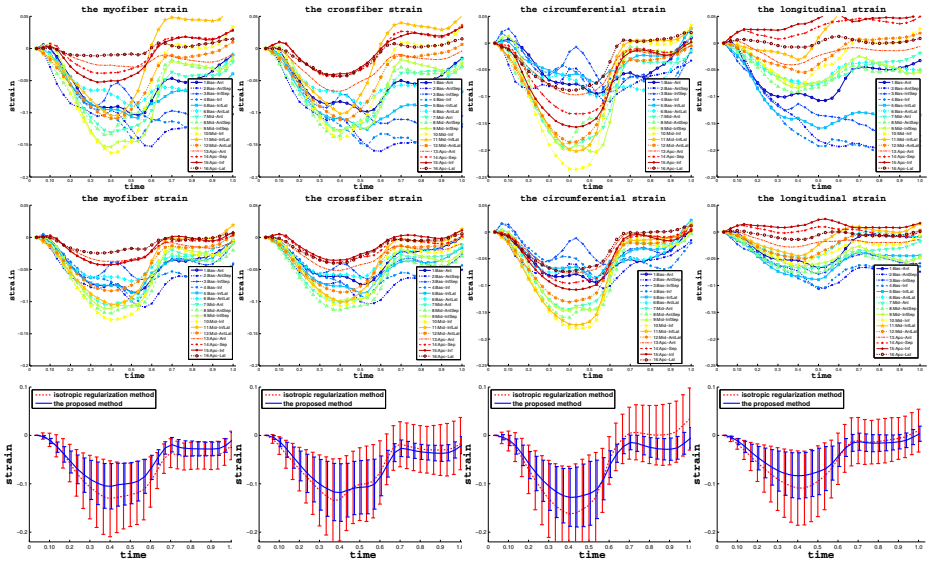

Fig. 3. The strains in the cine MRI sequence. (Top row) Strains in the isotropic regularization method. (Middle row) Strains in our method. (Bottom row) Average and transmural variance of the myofiber strains in four segments using isotropic regularization method (red) and our method (blue). Same segments are used as in Fig.2.

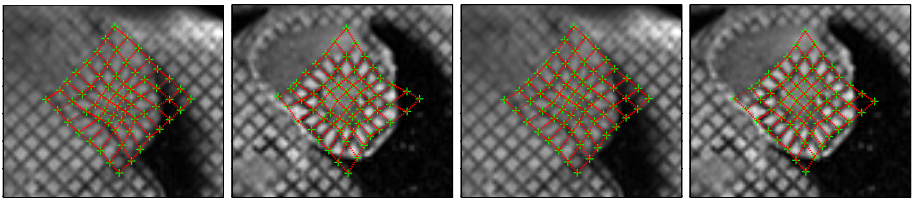

Fig. 4. The deformed ED frame grid overlaid on the ES frame in apex and basal slices in isotropic regularization method (left two) and our method (right two)

5 Conclusion

We proposed a diffeomorphic cardiac motion estimation method with transmurally myofiber strain homogeneity. We validated our method by using 3D echocardiography and MRI sequences of two healthy subjects. Experiments show that our method can obtain result which is more transmurally smooth and performs better in tracking myocardium points. Our method can be applied to pathological subjects whose myofiber alternation is not significant if given the pathological myofiber orientation. The method may have limitation in cases of the subjects with severe pathology such as infarct, in these cases, our method can be applied to the non-infarct regions.

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