Construction of Cardiac Anatomical Models Using Deformable Model Methods

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Abstract. Two deformable model methods based on the active surface and the active shape model methods are proposed in this paper for cardiac anatomical model construction. Several extensions to the original work [1,2] are introduced to deal with the segmentation of 4D cardiac images: The image force is calculated through gradient vector flow [3], which is also used in the landmarking for 3D triangle meshes. A multi-stage segmentation mechanism is proposed integrating the active surface and active shape model methods. This delivers better segmentation results than using either method alone.

1 Introduction

Cardiac electrophysiological and electromechanical modeling is one of the most important issues in mathematical modeling of the heart. Starting point for the modeling of the whole heart is a realistic anatomical model. In order to get the anatomical model, segmentation needs to be done on images captured through imaging modalities like MRI, CT, Ultrasound and etc. Cardiac segmentation is a difficult task because of the unavoidable presence of imaging noise and the intricate structure of the heart.

The active contour method [1] is now state of the art for medical image segmentation. By introducing high level shape based constraints, it offers better performance than the traditional low level methods like thresholding and region growing. The shape constraints introduced in the active contour method are based on the assumption that the deformable model shows the characteristics of an elastic body, which is the case in most of the medical image segmentation applications. However, this kind of constraint is too general and in some cases not sufficient to deliver satisfying results. To deal with this problem, the active shape model method [2] introduces more specific knowledge, namely the statistical shape model. The statistical shape model ensures the model to deform only in an allowable variation space derived from a training set, thus the difficulties resulting in undesirable deformation can be partly overcome. For cardiac image segmentation, the active shape model method is of particular advantage because the different chambers of the heart can be described effectively by a statistical shape model.

In this paper, the active contour and active shape model methods are realized with several extensions to the original methods. A novel landmarking method for triangle meshes is realized by deforming the meshes with the internal spring force and gradient vector flow image force [3]. The proposed multi-stage segmentation mechanism yields the best results, it utilizes the active shape model method in the first stage, followed by a relaxation stage as a refinement by using the active surface method. The mathematical backgrounds and realization procedures are explained in Sect. 2. Experimental results are shown in Sect. 3. The paper is concluded by Sect. 4 with some discussions and suggestions for further research.

2 Methods

The 2D active contour method [1] is extended to 3D active surface method in our project. Triangle meshes are used to represent the deformable surfaces. Evolution of the triangle meshes is calculated using a dynamic formulation of the original energy minimization concept. By the dynamic formulation, the deformation of the surface is regulated by the internal and external forces using the following Lagrange equation,

$$\mu \frac{\partial^2 s}{\partial t^2} + \gamma \frac{\partial s}{\partial t} + F_{\mathrm{internal}} + F_{\mathrm{external}} = 0 \tag{1}$$

where μ is the mass density, γ is the damping density, and s is the deformable surface. When the deformation of the surface reaches the stable state, $\frac{\partial^2 s}{\partial t^2} = \frac{\partial s}{\partial t} = 0$, the forces also reach their equilibrium state, $F_{\mathrm{internal}} + F_{\mathrm{external}} = 0$.

Corresponding to the surface expression using a triangle mesh, the internal force is computed as the combination of the internal stretching and bending forces as described in [1]. The external image force is calculated through a long range force, gradient vector flow (GVF) [3]. Compared with the traditional gradient image force, GVF has a much larger influence area, and presents itself even in homogeneous convex regions. In 3D space, GVF is defined as the vector field $\boldsymbol{V}(x, y, z) = [u(x, y, z), v(x, y, z), w(x, y, z)]$, and it is computed as the weighted diffusion of the image edge map f,

$$u_{\mathrm{t}} = \mu \nabla^2 u - (f_{\mathrm{x}}^2 + f_{\mathrm{y}}^2 + f_{\mathrm{z}}^2)(u - f_{\mathrm{x}}) \tag{2}$$

$$v_{\mathrm{t}} = \mu \nabla^2 v - (f_{\mathrm{x}}^2 + f_{\mathrm{y}}^2 + f_{\mathrm{z}}^2)(v - f_{\mathrm{y}}) \tag{3}$$

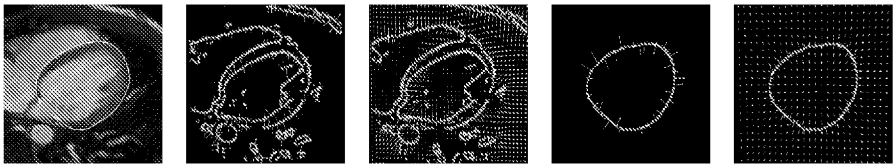
$$w_{\mathrm{t}} = \mu \nabla^2 w - (f_{\mathrm{x}}^2 + f_{\mathrm{y}}^2 + f_{\mathrm{z}}^2)(w - f_{\mathrm{z}}) \tag{4}$$

where μ is the weighting parameter, and governs the tradeoff between field smoothness and gradient approximation. The steady-state solution of the above parabolic equations is the desired GVF field. The Gauss-Seidel method with successive over relaxation (SOR) is used to compute the GVF field.

Following the methods proposed above, segmentation of 3D cardiac images using the active surface method is carried out using the following procedure:

1. Calculate the edge map from the original image using gradient operator;

Fig. 1. Comparison of the different image forces: (a) slice 53 of the original MRI image at heart phase 2 overlaid with the manually segmented surface mesh, (b) Sobel gradient force computed from the edge map of the original MRI image, (c) corresponding GVF force. (d) Sobel gradient force computed from the edge map of the manual segmentation result, (e) corresponding GVF force used for landmarking.



(a) (b) (c) (d) (e)

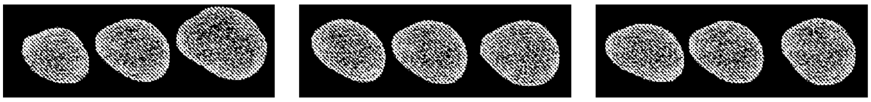
2. Calculate GVF from the edge map;
3. Construct an initial triangle mesh with the corresponding size and position to the object to be segmented;
4. Deform the initial triangle mesh with the combination of weighted internal force and GVF force until force equilibrium is reached. The final state of the triangle mesh is the segmentation result.

For segmentation using the active shape model method, the first step is to construct the statistical shape model from the manually segmented training set. Manual segmentation is based on three-dimensional mesh warping technique. After manual segmentation, the triangle meshes need to be landmarked to establish correspondence between vertices. A natural way for landmarking is to use anatomical landmarks, but for 3D cardiac images, the anatomical landmarks are hard to find and are not enough to fully express the shape. A modified active surface method is used in our system to perform landmarking. This is based on the method proposed in [4], however the force equilibrium method is used in our method to deform the average template mesh instead of the energy minimization method. GVF force is used in the landmarking process, because experimental results show that GVF force offers a much better performance for the active surface based landmarking. The reason for this is that GVF algorithm generates smoother and broader force fields from the segmented images than the simple gradient operator.

After landmarking, principle component analysis (PCA) is used to derive the average shape and the allowable variation space from the landmarked triangulation meshes. This statistical shape model is then used to regulate the segmentation process in a recursive way,

1. Set the initial triangle mesh to the average shape of the statistical shape model.
2. Set $\frac{\partial^2 s}{\partial t^2} = 0$, then (1) is converted into: $\gamma \frac{\partial s}{\partial t} + F_{\text{internal}} + F_{\text{external}} = 0$. Compute the intermediate surface s using this equation with a user adjustable step length.

Fig. 2. Variations of left ventricle epicardium shape model corresponding to the three largest eigenmodes (a-c).



(a) (b) (c)

3. Calculate the projection parameters of the intermediate surface s to the statistical shape model. Truncate the projection parameters when they are outside the allowable shape space.
4. Reconstruct the intermediate surface s into $\tilde{s}$ using the projection parameter.
5. Set $\tilde{s}$ as the initial shape, repeat the whole process until there are no obvious changes of the surface $\tilde{s}$.
6. Refine $\tilde{s}$ by deforming it within a small neighbourhood region (e.g. $3 \times 3 \times 3$) using active surface method proposed above in this paper.

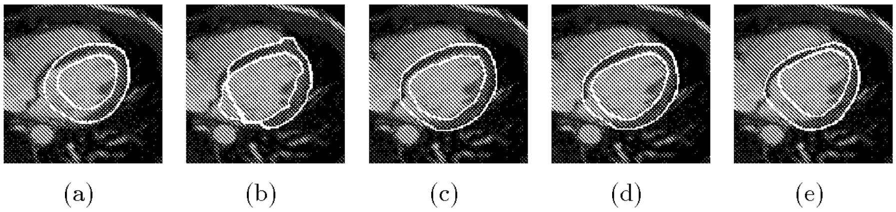
3 Results

To test the algorithms and compare the results of different methods, a 4D MRI data set is used. This data set is acquired through a balanced FFE sequence (TrueFISP), and contains 25 phases of a cardiac cycle. The statistical shape model is constructed from a training set formed by 20 heart phases ($6 \sim 25$). Then the shape model is used as a regulating factor to segment other heart phases ($1 \sim 5$) using the active shape model method presented in the above section, and the results are compared with the manual and the active surface methods. Fig. 1 shows the force distributions of the original cardiac image and the manually segmented left ventricle epicardium. It is shown that GVF forces have broader and smoother influence area. This characteristic of GVF decreases the sensitivity to the model initialization and the surface evolution under GVF is smoother and faster. Variations of the left ventricle epicardium shape model are shown in Fig. 2. From the PCA results, the first 7 parameters represent more than 90% of the variation. The segmentation results are shown in Fig. 3. It is shown that the active shape model method combined with the active surface method for final relaxation offers the best results among the automatic methods proposed in this paper, and is comparable in quality to the manual segmentation method. For segmentation of 4D cardiac images, the result of one heart phase can be used as the initial shape for the following heart phase.

4 Discussion

In this paper, a modified active shape model method is presented for construction of cardiac anatomical model. The statistical shape model serves as a new

Fig. 3. Segmentation results of the left ventricle epi- and endocardium for slice 53 of heart phase 1: (a) Initial surfaces, (b) result using the active surface method, (c) result using the active shape model method before relaxation, and (d) after relaxation, (e) manual segmentation result.



regulating factor that prevents the surface model from deforming into undesirable regions under the influence of image force. Subjective evaluation ("visual inspection") shows the better performance of the modified active shape model method for the tested 4D MRI data set. For more general conclusions, more data sets need to be tested, and a more objective evaluation method has to be defined.

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