3-D deformable model for aortic aneurysm segmentation from CT images

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Abstract— **For treatment of abdominal aortic aneurysm (AAA) by placement of aortic stent graft device it is necessary to make accurate AAA measurements in order to choose the stent graft device of appropriate shape and size. In this paper, we propose a novel technique for 3-D segmentation of abdominal aortic aneurysm from computed tomography (CT) angiography images. The technique is based on 3-D deformable model and utilizes the level set algorithm for implementation of the method. The method performs 3-D segmentation of CT images and extracts a 3-D AAA model. Once the 3-D model of AAA is available it is easy to perform all required measurements for appropriate stent graft selection. The method proposed in this paper uses the level-set algorithm instead of the classical active contour algorithm developed by Kass et al. The main advantage of the level set algorithm is that it enables easy segmentation of complex structures such as bifurcations in arteries. In the level set approach for shape modeling, a 3-D surface is represented by a real 3-D function that can be viewed as a 4-D surface. The 4-D surface evolves through an iterative process of solving the differential equation of surface motion. The surface motion is defined by velocity at each point. The velocity is a sum of a constant velocity (inflation force), curvature-dependent velocity (internal force), and image-dependent velocity (external force). The image-dependent velocity is computed on the basis of image gradient. The algorithm has been implemented in MATLAB and C languages. Experiments have been performed using real patient CT angiography images and have shown good results. A 3-D rendering of the segmented region is performed that is useful for aneurysm shape visualization.**

Keywords— **aortic aneurysm, aorta, image analyis, deformable models, level set algorithm**

I. INTRODUCTION

Abdominal aortic aneurysm (AAA) [1], [2] is a cardiovascular disease that can be treated by open surgery and by placement of aortic stent graft. The main disadvantage of open surgery treatment is the invasive nature of the procedure. For this reason, an alternative procedure has been proposed recently that is based on endovascular placement of aortic stent graft through a minimally invasive opening on the patient body. The main advantage for the patient is that stent graft device requires a less invasive procedure. However, the difficulty with this method is that accurate measurements of AAA must be made prior to the surgical procedure in order to choose the stent graft device of appropriate shape and size. A number of different modalities have been used for imaging of AAA, such as ultrasound, digital subtraction angiography (DSA), CT angiography, and contrast-agent enhanced MR angiography [3]. Modern medical imaging techniques followed by appropriate image analysis methods [4] have shown to be useful for measurements of AAA [5], [6].

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In this paper we describe a 3-D image analysis technique for segmentation of AAA from CT angiography images. The technique is based on 3-D deformable model and utilizes a level set algorithm for implementation of the method. The method performs 3-D segmentation of CT images and extracts a 3-D AAA model. Once the 3-D model of AAA is available it is easy to perform all required measurements for appropriate stent graft selection.

In our previous work [6], we have developed a semi-automatic 3-D technique for aneurysm segmentation that is based on 2-D active contours with additional forces introduced for 3-D interaction between the slices. The method uses the clasical active contour algorithm developed by Kass et al. [7]. The method proposed in this paper uses the level-set algorithm [10] instead of the classical active contour algorithm. The main advantage of the level set algorithm in our application is that it enables easy segmentation of bifurcations in arteries.

The organization of the paper is as follows. The level set method for deformable model-based segmentation is described in Section II. The proposed method for AAA segmentation is described in Section III. Results and discussion are presented in Section IV. The conclusion is provided in Section V.

II. LEVEL-SET METHOD FOR DEFORMABLE MODEL-BASED SEGMENTATION

Deformable models have shown to be a powerful tool for medical image segmentation [8]. The original active contour algorithm [7] uses active contours (snakes). The classical snake approach has several disadvantages: (i) difficulties with segmentation of topologically complex structures, and (ii) complex implementation in 3-D. To overcome these difficulties, the level set method has been proposed [9], [10]. In this approach for shape modeling, a 2-D curve γ is represented by a 3-D surface Ψ. The value of the 3-D surface at some point x is defined as a distance d of the point x to the 2-D curve according to Equation 1.

$$\Psi(x, t = 0) = \pm d \tag{1}$$

where $x \in R^2$ are points in image space. The sign determines whether the point lies outside or inside the 2-D curve $\gamma(t = 0)$. In this manner, the 2-D curve is represented by the zero level set $\gamma(t) = \{x \in R^2 \mid \Psi(x, t) = 0\}$ of the level set function. The level set method then evolves the 3-D surface instead of the original 2-D curve. The motion of the 3-D surface is described by means of a partial differential equation (PDE) shown in Equation 3. This PDE is solved using entropy-satisfying schemes borrowed from the numerical solution of hyperbolic conservation laws which produce the correct viscosity solution [10].

While making things more complex, the level set method introduces some new qualities and resolves some problems found in the classical snake method. As long as the surface stays smooth, its zero level set can take any shape and even change topology. Another advantage is that it is easy to build accurate numerical schemes to approximate the equations of motion. Curvatures and normals may be easily evaluated and the technique extends trivially to higher dimensions.

The evolution PDE for evolution of the function $\Psi(x,t)$ has the following form:

$$\frac{\partial\Psi(x,t)}{\partial t} + F\left|\nabla\Psi\right| = 0 \tag{2}$$

with a given initial condition $\Psi(x, t = 0)$.

For numerical solution of the Equation 3 it is necessary to perform discretization in both space and time domains. For this purpose we discretize space coordinates using a uniform mesh of spacing h, with grid nodes denoted by indices ij. Let Ψ_{ij}^{n} be the approximation to the solution $\Psi(ih, jh, n\Delta t)$, where Δt is the time step. The expression for Ψ_{ij}^{n+1} can be derived using the upward finite difference method:

$$\frac{\Psi_{ij}^{n+1} - \Psi_{ij}^{n}}{\Delta t} + F\left|\nabla_{ij}\Psi_{ij}^{n}\right| = 0 \tag{3}$$

which gives us the final iteration expression:

$$\Psi_{ij}^{n+1} = \Psi_{ij}^{n} - \Delta t F\left|\nabla_{ij}\Psi_{ij}^{n}\right| \tag{4}$$

The speed term F depends on the curvature K and is separated into a constant advection term F_0 and the remainder $F_1(K)$, that is

$$F(K) = F_0 + F_1(K) \tag{5}$$

The advection term F_0 defines a uniform direction speed of front which would be achived at regions of front with zero curvature. The difusion term $F_1(K)$ depends on the local curvature and smoothes out regions of high curvature. We use the following expression for the speed term

$$F = 1 - \epsilon K \tag{6}$$

where ϵ is the entropy condition which regulates the smoothness of the curve. The proposed range for ϵ is 0.5 to 1.0.

The curvature is obtained from the divergence of the gradient of the unit normal vector to front, that is

$$K = \nabla\cdot\frac{\nabla\Psi}{|\nabla\Psi|} = \frac{\Psi_{xx}\Psi_y^2 - 2\Psi_x\Psi_y\Psi_x y + \Psi_{yy}\Psi_x^2}{(\Psi_x^2 + \Psi_y^2)^{3/2}} \tag{7}$$

The speed function F also has to have an image based condition which would cause propagating front to stop in the vicinity of desired object boundary. This image term is defined at zero level set only and has to be extended over all level sets. We include influence of image gradient by multiplying the speed function F with a quantity k. The term k can be defined in many ways and we use the following form.

$$k(x,y) = e^{-|\nabla G_\sigma * I(x,y)|} \tag{8}$$

where $G_\sigma * I$ denotes image convolved with Gaussian smoothing filter whose characteristic width is σ. We use a narrow band extension as proposed by Sethian et al. [10] where front is moved by updating the level set function at a small set of points in the neighborhood of zero level set called the narrow band, instead of updating it at all points on the grid. Two boundary curves of the narrow band are δ apart (level sets $\Psi = \pm\delta/2$). During a given time step the value of Ψ outside the narrow band is stationary and propagating front cannot move past the narrow band. After a given number of iterations the curve γ, the level set function, and the new narrow band are recalculated. Using such an approach the image based term needs to be calculated only in the narrow band and can be calculated for each Ψ point based on its corresponding Gaussian gradient point.

In the above text, we have described the level set method for two dimensional images. Extension to three dimensions is straight forward by extending the array structures and gradient operators. In that case, Ψ is a 4-D surface and we use the following expression for curvature of the level set function as stated in [9].

$$\begin{aligned} K &= \frac{\Psi_{xx}(\Psi_y^2+\Psi_z^2) + \Psi_{yy}(\Psi_x^2+\Psi_z^2) + \Psi_{zz}(\Psi_x^2+\Psi_y^2)}{(\Psi_x^2+\Psi_y^2+\Psi_z^2)^{3/2}} \\ &- \frac{2\Psi_x y\Psi_x\Psi_y + 2\Psi_x z\Psi_x\Psi_z + 2\Psi_y z\Psi_y\Psi_z}{(\Psi_x^2+\Psi_y^2+\Psi_z^2)^{3/2}} \end{aligned} \tag{9}$$

III. 3-D ABDOMINAL AORTIC ANEURYSM SEGMENTATION

The above described 3-D level set method is applied to the problem of AAA segmentation. The input to the level set algorithm in this case is a 3-D data array of volumetric CT angiography data of the human abdomen. The region of interest containing the AAA is manually extracted from the volumetric data. This step does not have any influence on execution of the algorithm but greatly reduces the memory costs. In order to use the level set algorithm, an initial surface has to be defined. We choose the initial surface to be a sphere. The sphere center and radius have to be defined manually by the user. The initial sphere has to be placed so that it resides entirely inside the abdominal aorta. The algorithm described below, then evolves the initial surface until it stops changing. The outline of the procedure is shown in Figure 1. The output of the

Define initial surface γ.
Calculate initial Ψ.
repeat
    for i=1, $\ldots$, N_{iter} **do**
        Execute iteration in Equation 4
    end for
    Recalculate surface γ
    Recalculate narrow band
    Reinitiate Ψ in narrow band
until γ stops changing

Fig. 1. The level set algorithm for abdominal aortic aneurysm segmentation.

algorithm is the final surface γ representing the internal boundary of the aorta.

The major part of the algorithm has been implemented in MATLAB program package while the most computationally complex steps of reacalculating γ and narrow band and reinitialization of Ψ, have been implemented in C programing language.

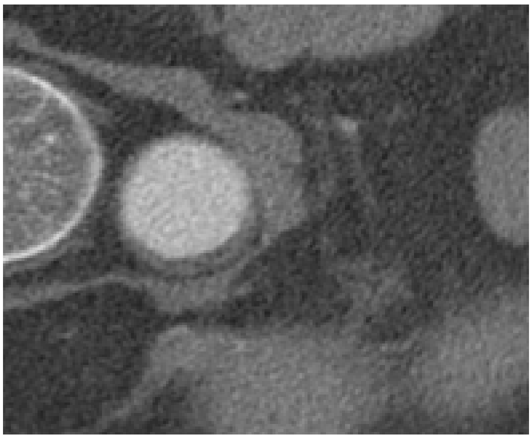
IV. RESULTS AND DISCUSSION

The algorithm has been tested using CT angiography images of a real patient. The CT volume consists of 100 images of dimension 350×450. Segmentation has been performed on a manually pre-selected region of interest around aorta. The resolution within the region of interest is reduced in x and y dimensions by a factor of 2. Values of numerical constants are as follows. The constant appearing in the speed term 7 is set to $\epsilon = 0.9$ so that lateral branching of aorta is not segmented. Gaussian smoothing filter characteristic width $\sigma = 0.9$ and Gaussian kernel width 7 has been used. The constant N_{iter} which determines the number of inner loops in algorithm in Figure 1 is set to 4 which is influenced by the width of narrow band $\delta = 6$. The gradient of input data in Equation 8 produced by convolving the input data with derivative of Gaussian kernel has magnitude of 20 at aorta border which is not enough to stop Ψ motion so the gradient value had to be multiplied by a factor of 100.

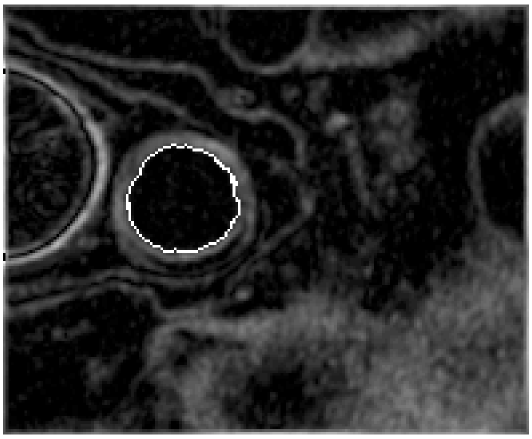
Figures 2 and 3 show the results of segmentation using the proposed algorithm. Subfigure a) in both figures shows a slice of input data, subfigure b) shows the computed gradient of the corresponding slice with the resulting curve and subfigure c) shows the resulting curve superimposed on the original slice. Three dimensional model of resulting surface is shown in Figure 4.

V. CONCLUSION

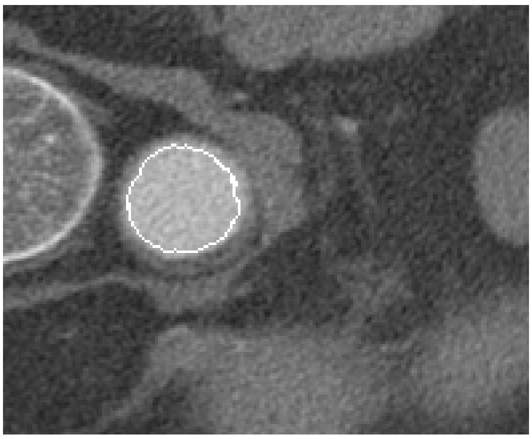
Stent graft placement has been recently introduced as a method for less invasive treatment of abdominal aortic aneurysm. This techniques requires accurate measurements of the aneurysm for selection of appropriate stent graft shape and size. These measurements are performed by imaging the patient using various medical imaging modalities. In this paper, we have presented a novel 3-D algorithm for abdominal aortic aneurysm segmentation



(a) Original image.

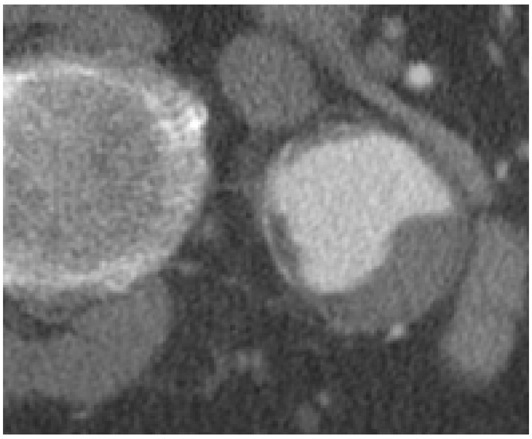


(b) Gradient image with contour.

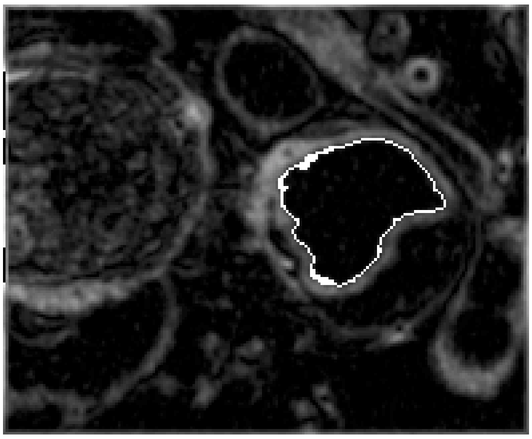


(c) Original image with contour.

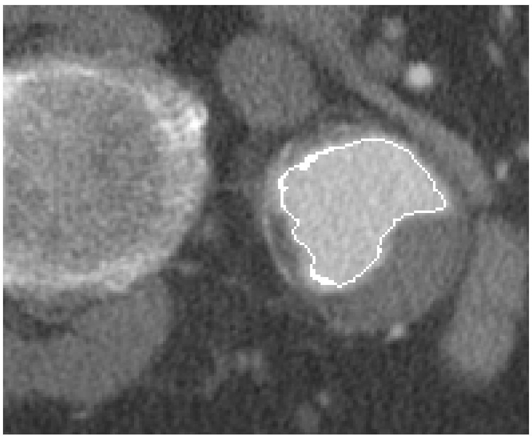
Fig. 2. Segmentation results of abdominal aorta at slice 11.



(a) Original image.



(b) Gradient image with contour.



(c) Original image with contour.

Fig. 3. Segmentation results of abdominal aorta at slice 43.

Fig. 4. 3-D visualization of segmented abdominal aorta.

from CT images. The algorithm uses a 3-D deformable model and is implemented using the level set algorithm.

Experiments have been performed using CT images of patients having abdominal aortic aneurysm. Experiments have shown good results.

Future work will include investigation of the influence of variation of the initial surface to numerical convergence speed and different approaches to gradient computation.

References

[1] C. B. Ernst, "Abdominal aortic aneurysms," *New England Journal of Medicine*, vol. 328, pp. 1167–1172, 1993.

[2] R. M. Berne and M. N. Levy, *Cardiovascular Physiology, 7th Ed.*, Mosby-Year Book, 1997.

[3] S. A. Thurnher, R. Dorffner, M. M. Thurnher, F. W. Winkelbauer, G. Kretschmer, P. Polterauer, and J. Lammer, "Evaluation of abdominal aortic aneurysm for stent-graft placement: Comparison of Gadolinium-enhanced MR angiography versus helical CT angiography and digital subtraction angiography," *Radiology*, vol. 205, pp. 341–352, 1997.

[4] A. P. Dhawan and S. Juvvadi, "Knowledge-based analysis and understanding of medical images," *Computer Methods and Programs in Biomedicine*, vol. 33, pp. 221–239, 1990.

[5] M. Garreau, J. L. Coatrieux, R. Collorec, and C. Chardenon, "A knowledge-based approach for 3-D reconstruction and labeling of vascular networks from biplane angiographic projections," *IEEE Transactions on Medical Imaging*, vol. 10, pp. 122–131, 1991.

[6] S. Loncaric, D. Kovacevic, and E. Sorantin, "Semi-automatic active contour approach to segmentation of computed tomography volumes," in *Proceedings of SPIE Medical Imaging*, 2000, vol. 3979, p. to be published.

[7] M. Kass, A. Witkin, and D. Terzopoulos, "Snakes: active contour models," *International Journal of Computer Vision*, vol. 1, pp. 321–331, 1987.

[8] T. McInerey and D. Terzopoulos, "Deformable models in medical image analysis: A survey," *Medical Image Analysis*, vol. 1, pp. 91–108, 1996.

[9] S. Osher and J. A. Sethian, "Fronts propagating with curvature dependent speed: algorithms based on hamilton-jacobi formulation," *Journal of Computational Physics*, vol. 79, pp. 12–49, 1988.

[10] R. Malladi, J.A. Sethian, and B.C. Vemuri, "Shape modeling with front propagation," *IEEE Transactions on PAMI*, vol. 17, pp. 158–176, 1995.