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Generation of structure of the aortic bifurcation from magnetic resonance angiogram

Q. Long, X.Y. Xu, M.W. Collins
Thermo-Fluids Engineering Research Centre, City University,
London EC1V 0HB, UK

Abstract

Magnetic Resonance Imaging (MRI) can produce a series of parallel 2D cross-sectional images of an arterial vessel. Based on these images, the three-dimensional structure of such a vessel can be obtained by image processing techniques. In this paper, a novel image processing and three-dimensional reconstruction approach is presented which permits the recreation of the aortic bifurcation from magnetic resonance angiograms. The reconstruction output not only provides 3D visualisation of the bifurcation structure, but also serves as an interface for further quantitative analysis of the fluid dynamics in the model.

1 Introduction

Although it is now recognised that specific patterns of local blood flow predispose to the development of atheroma, the mechanisms underlying this important determinant of cardiovascular "risk" are incompletely understood. Atheroma develops preferentially where arteries branch, and the human aorto-iliac bifurcation provides an ideal model for studying the influence of branching geometry. This is because the incidence of disease at this location is related to branching asymmetry, and thus to different patterns of flow in the common iliac arteries.

In the past, due to the complex geometric structure of the aortic bifurcation, it is difficult to obtain an anatomically realistic model for detailed study of the flow in the branch. Recently however, Magnetic Resonance

Angiography (MRA) techniques have made this possible. MRA can present diagnostic vascular information as an alternative to existing noninvasive methods without the need for a contrast material. Scan sequences based on either the "time-of-flight" effect or "phase shift" effect generate a series of two-dimensional images of the arterial vessel and their spacial relationship. From these 2D cross-sectional images which are usually parallel to each other, three-dimensional structures of the arterial vessel can be obtained by using appropriate techniques for further data/image processing.

The processed 3D data set contains quantitative information of the structure of vessel, in this case, the aortic bifurcation. It may be read directly by CAD/CAM softwares for construction of physical casts [1], in which haemodynamic experiments are performed. Furthermore, by means of our purpose-built image processing software, it may serve as an interface for further modelling using Computational Fluid Dynamics (CFD).

Three-dimensional structural reconstruction is usually performed first by reading in a series of two-dimensional images, and then generating either surface patches which describe the *boundary* of the object or a volume which describes the *structure* of the object. These latter are known as surface and volume rendering respectively. Many methods have been developed to reconstruct 3D geometries. Boissonnat[2] used the volume rendering method to generate a 3D structure from serial planar contours, while Chen et al[3] applied the surface rendering approach to generate the geometry of the bladder. Cohen and Cohen[4] developed a balloon model for recreation of 3D active surfaces from MR images, and the Marching Cubes algorithm was employed by Lorensen and Cline[5]. Although these approaches are capable of generating high quality 3D images, they are complicated and cannot guarantee a satisfactory result if the 2D images are of poor quality. In this paper, we propose a novel 3D geometrical reconstruction method which was developed in our centre, Long et al[6]. The method is robust and able to produce both a reliable 3D structure and a smooth surface of an arterial vessel. The resultant 3D data set has been organised in a format that allows for automatic numerical grid generation to be performed, which is essential for subsequent CFD studies.

2 Method

The generation of a 3D structure from an angiogram generally involves the following steps: 1) the acquisition of 2D cross-sectional images, 2) the definition of 2D contours by image processing, 3) 3D reconstruction, and 4) the visualisation of the 3D structure. In our approach an extra step 5) was carried out to reorganize the 3D data set for numerical grid generation.

Typical MR angiograms of the aortic bifurcation were acquired in two subjects. The MR imaging was performed by a 1.5T scanner (Signa, GE medical systems), using a 2D time-of-flight gradient echo sequence (TR 45 ms, TE 8.7ms, Flip Angle 60, NEX=1) and contiguous 1.5mm slices with an in-plane resolution of 1.41mm. The scan covered the region from 10cm proximal to the abdominal bifurcation to 15cm distal to the bifurcation. These images were transferred for storage and manipulation via a local network using XFER (GE Medical Systems), the data transfer program, from the scanner's computer to a Sun workstation.

Once the parallel images became available at the workstation, preprocessing steps were performed. As it was very desirable to avoid loss of details of the images near the bifurcation region, only a small size Gaussian convolution was performed to smooth the image in order to obtain a clear and continuous edge of the vessel. One of the most important steps in 3D angiogram generation is to produce high quality contours of the vessel, in other word, to make a good segmentation from the 2D images. There are a number of segmentation techniques available in image processing, with regard to the problem considered here, because the arterial vessel normally has a smooth intro-surface. This means the segmentation method should consider not only image intensity distribution but also the smoothness of the contour. One of the most suitable methods is the active contour model which was first introduced by Kass et al[7]. The active contour or *snake* model considers a deformable contour as possessing internal energy in order to impart smoothness to the contour. When this contour is acted on by an external energy field, the contour seeks equilibrium at a local minimum of the field by moving and changing shape in a 'snakelike' manner. The external energy field provides a flexible mechanism to incorporate information about the boundary to be extracted. In this paper such a snake model was employed for the 2D segmentation.

The model used here includes a number of control points connected by series of B-splines. The energy definitions are:

$$\mathbf{E} = E_{inter} + E_{exter} \tag{1}$$

where E_{inter} is the internal energy and E_{exter} is the external energy,

$$E_{inter} = \int (w_{cont} E_{cont}(s) + w_{curv} E_{curv}(s)) ds \tag{2}$$

$$E_{exter} = \int (w_{int}(s) E_{int}(s) + w_{edge}(s) E_{edge}(s)) ds \tag{3}$$

where E_{cont} corresponds to the tangent of the contour at the control point, E_{curv} is the curvature, E_{int} represents the image intensity term of the energy and E_{edge} the edge term of the energy. s corresponds to a special location on the contour (here is the control points). The integral performs along the contour. A higher value of E_{inter} tends to smooth the contour and a higher value of E_{exter} allows the contour to be attracted more easily towards the edge of the vessel. Therefore the ratio of these two terms reflects the

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smoothness/closeness of the contour.

To perform the *snake* model, then, an initial guess of the contour should be provided, for which a region growing method was used. As the initial contour obtained was enclosed by edge points one by one (boundary of the region), a down size step was performed to make several pixels gap between each control point. The internal and external energy were calculated at each control point and at the pixels adjacent to the control point along the normal direction of the contour. The local energy minimum points were chosen as the new control points to be used in the next iteration. The convergence can be monitored by calculating the global energy $\mathbf{E}$, which should become increasingly smaller with each iteration. If the distance moved by the control points is smaller than the given tolerance, the calculation should converge. We have found the initial contour calculated using the region growing method is very close to the final solution, convergence can normally be achieved in less then 10 iterations.

Having obtained a series of contours, the 3D reconstruction of the aortic bifurcation was achieved by a three-dimensional reconstruction approach specially developed in our centre, Long et al[6]. The algorithm involves the performance of surface smoothing and interpolation steps. Because MR angiography of the abdominal aorta always has artifacts due to physiological cardiac or respiratory motion [8], a compensation step is necessary in some cases. The ratio of the smoothness/closeness of the surface can be easily controlled by a parameter γ. The end result of the reconstruction is a structured surface points data file. To visualise the structure surfaces are formed encompassing these points using surface interpolation if the resolution is not satisfied. To prepare the data file for grid generation, B-spline interpolations are performed in all directions. With this information either *structured* or *unstructured* grids can be generated.

3 Results

The progressive results of the two-dimensional image processing are shown in Fig.1. Fig.1.a gives the original MR image before any processing, in which the signal intensity is not uniform in the vessel; the location of this section is 4.5mm above the bifurcation. Fig.1.b shows the image after processing by a Gaussian smooth with $\sigma = 1.0$, and Fig.1.c shows the edge image obtained by using the sobel operator on the image in Fig.1.b. This edge image would be used in the snake model in calculating E_{edge} (pixel value corresponding to the gradient magnitude). Fig.1.d displays the edge of the vessel obtained by the region growing method performed on image Fig1.b, this edge being used as an initial contour in the snake model. The final contour generated by the snake model is given in Fig.1.e. Because these images have structured data format, the contour points can only have

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integer x,y coordinates. Infact the contour point coordinate can actually be real giving a smoother result than in the image shown;

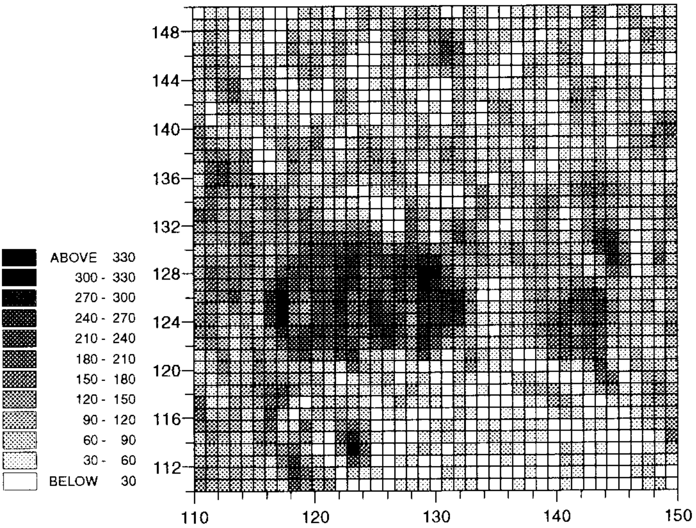

Figure 1.a: Original MR cross-sectional image. Unit: 1=1.41mm.

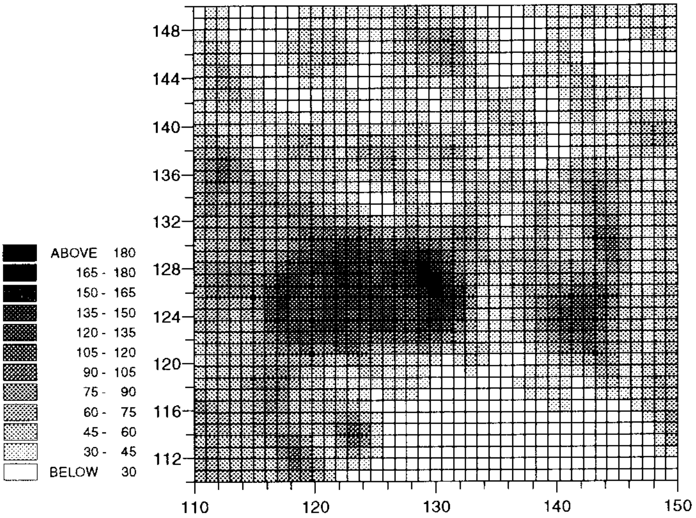

Figure 1.b: Smoothed image by Gaussian filter, $\sigma = 1.0$.

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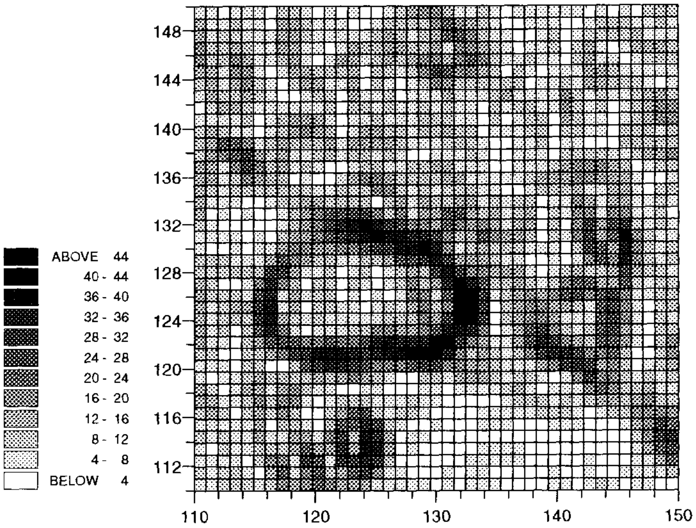

Figure 1.c: Edge image by Sobel operator from Gaussian image.

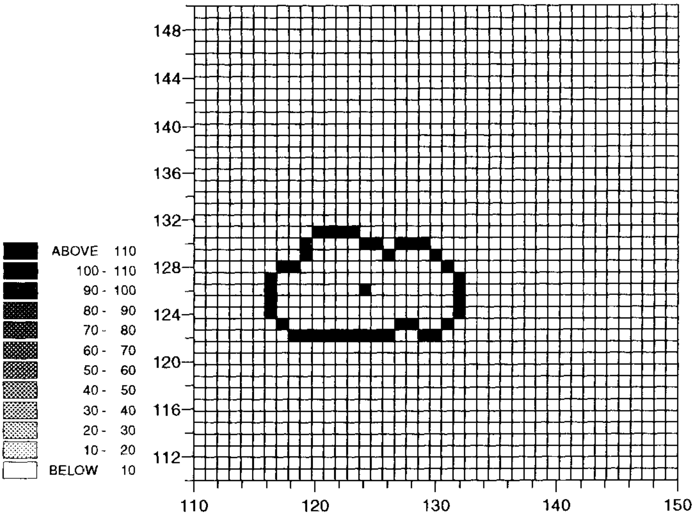

Figure 1.d: Vessel boundary by region growing, from the Gaussian image.

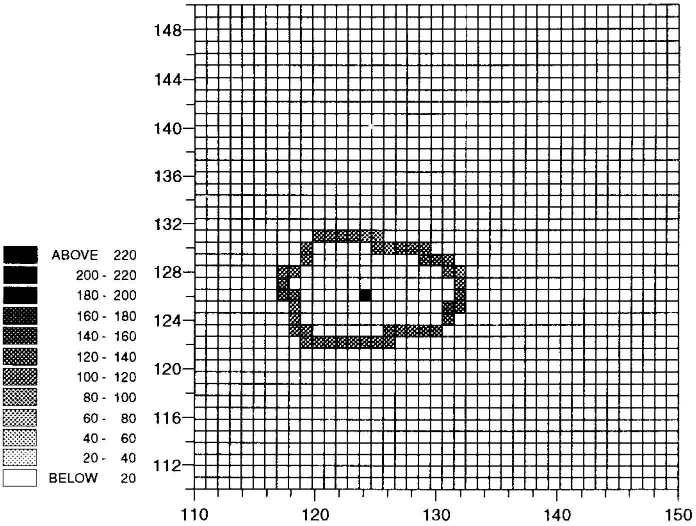

Figure 1.e: Final contour by the snake model.

Three-dimensional reconstruction results are displayed in Fig.2. Fig.2.a is the wire-frame surface of the aortic bifurcation with a smoothness parameter $\gamma = 1.0$. The effect of γ is shown in Fig.2.b which is as for Fig.2.a but with $\gamma = 2.5$. Fig.2.c is the shaded 3D surface view of the aortic bifurcation.

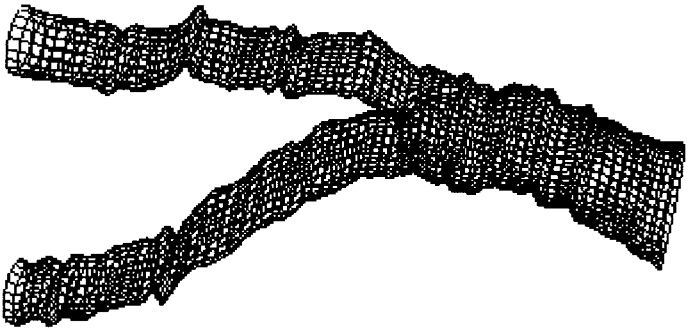

Figure 2.a: Wire-frame surface of aortic bifurcation with $\gamma = 1.0$

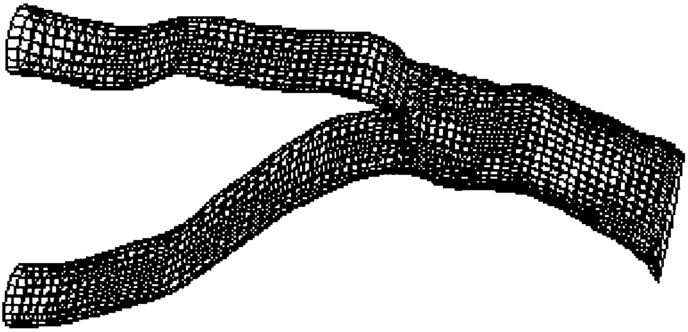

Figure 2.b: Wire-frame surface of aortic bifurcation with $\gamma = 2.5$

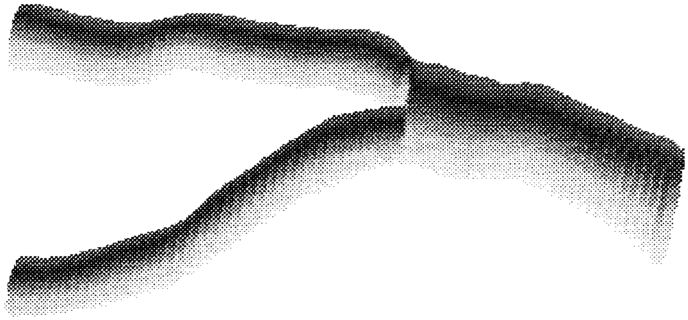

Figure 2.c: 3D surface image of aortic bifurcation with $\gamma = 2.5$

4 Discussion

Recently, numerous three-dimensional medical imaging methods utilizing hardware and/or software options have become commercially available. However most of these products are aimed at generating the surface of the object for visulization only; in other words, the surface rendering is the end product of most approaches. They are not suitable for further quantitative analysis of the 3D data set, which is very important in our study as the ultimate objective is to be able to use the reconstructed three-dimensional bifurcation geometry in our flow modelling. Results shown above have demonstrated that our approach can generate reliable 3D structures of the aortic bifurcation and the final surface points are very well organised and reusable for further flow studies.

In order to acquire more detail of the structure near the bifurcation region, thin slices of the MR angiogram are necessary. Therefore special precautions have to be taken when pre-processing images in this region. In this paper, only Gaussian smoothing has been used. It appears better than spatial filters such as median filter and nonlinear mean filter, Pitas et al [9]. A good edge detection method can be performed on this Gaussian image.

More than several hundreds of 2D MR images need to be processed to reconstruct a 3D aortic bifurcation. Should the time dependent arterial wall movement be taken into account, several thousands of images would have to be handled. An efficient and robust 2D image processing algorithm is crucial for this problem. In our approach, the region growing method requires input of seed point location and pixel intensity threshold. The software has been designed in a way that when processing serial images (to generate a segment of the vessel) the seed point are transfered from the first image until the last image. If the quality of the images is high, one pixel intensity threshold would be sufficient as the segmentation result is not very sensitive to the threshold. However, if the image intensity at the region of interest (ROI) varies significantly from one slice to another, individual thresholds would be necessary for each slice. In this case, an automatic threshold searching program woould be involved to find a suitable threshold. The snake model used here has proved to be very stable and without the need for operator interaction. Therefore, the above 2D procedure may fairly be described as *fully automatic.*

The three-dimensional reconstruction method we have developed is very easy to operate and can be applied to any cavity organ such as ventricle, bladder and arterial vessels. Only one operating parameter needs to be input manually, which is the smoothness factor γ. At present, our method has the following limitations: 1) the possibility of oversmoothing the surface due to inadequate number of sections (MR slice images) causing lumenal narrowing to be missed, especially if the slices are thick; 2) extra steps need to be performed for complex geometrical structures, (for example, to obtain the apex of the bifurcation, a special interpolation procedure would be required if the image slices are not sufficiently thin. In addition, the quality of the three-dimensional image is dependent directly on the corresponding quality of the originally acquired images. A high contrast between vessel lumen and background (e.g., “white blood” images), the same pixel intensity level at each slice, thin contiguous image slices, and minimization of motion artifacts would all help to produce superior 3D reconstructions.

In conclusion, the image processing and 3D reconstruction approach employed here is capable of generating 3D vascular models from 2D cross-sectional MR angiograms. The method has proved to be efficient, robust and reliable. The geometrical model of the aortic bifurcation obtained can be used for further analysis in the form of Computational Fluid Dynamics.

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Key words: Aortic Bifurcation, Magnetic Resonance Angiogram, Image Processing, Three-dimensional Reconstruction.

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