Simulation Environment for the Evaluation of 3D Coronary Tree Reconstruction Algorithms in Rotational Angiography.

Guanyu Yang, Alexandre Bousse, Christine Toumoulin, Huazhong Shu

To cite this version:


HAL Id: inserm-00188485
http://www.hal.inserm.fr/inserm-00188485
Submitted on 20 Nov 2007

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L’archive ouverte pluridisciplinaire HAL, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d’enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.
Simulation Environment for the Evaluation of 3D Coronary Tree Reconstruction Algorithms in Rotational Angiography

G. Yang¹,²,³, A. Bousse¹,²,³, C. Toumoulin²,³ and H. Shu¹,³
1. Laboratory of Image Science and Technology, Southeast University, Nanjing, 210096, P.R. China
2. INSERM U642, Laboratoire Traitement du Signal et de l’Image, Université de Rennes 1, Rennes, 35042, France
3. Centre de Recherche en Information Biomédicale Sino-français (CRIBS)

Abstract—We present a preliminary version of a simulation environment to evaluate the 3D reconstruction algorithms of the coronary arteries in rotational angiography. It includes the construction of a 3D dynamic model of the coronary tree from patient data, the modeling of the rotational angiography acquisition system to simulate different acquisition and gating strategies and the calculation of radiographic projections of the 3D model of coronary tree throughout several cardiac cycles.

Index Terms—Coronary Artery, 3D Reconstruction, Geometric and Physic Models, 3D Rotational Angiography

I. INTRODUCTION

Percutaneous coronary interventions have grown over these last years to become a common revascularization method for the treatment of coronary artery disease.

Even if the traditional biplane X-ray angiography remains the gold standard for both diagnosis and image guided treatment, it has been shown unable to provide reliable geometric information on the lesion (degree of stenosis, length, eccentricity, diameter of the vessel and plaque morphology). It often involves to multiplying the incidences to find the optimal views which will show the lesion clearly while minimizing vessel overlap and foreshortening of vessel segments in areas of interest. The development of the 3D rotational angiography (3D-RA) system brings new perspectives for the 3D reconstruction of the coronary tree. The rotational image acquisition allows the cardiologist to obtain up to 180 projections of the left or right coronary tree during a single injection of contrast under different angles (caudal, cranial, axial). Today, the challenge is to exploit the set of available projections to perform 3D reconstruction of the coronary tree taking into account two kinds of movements: the non linear motion of the structures and the rotation of the acquisition system.

Different techniques are considered for the reconstruction of the coronary tree from projections. A first class relies on 3D modeling techniques either from two views[1] or from selected projections at the same cardiac phase (or rest phase) when considering a rotational acquisition [2]. These methods exploit the epipolar geometry to achieve the 3D reconstruction of coronary tree. A second category used a pre-computed motion model to modify the projection operator and calculate a motion compensated 3D tomographic reconstruction based on ART[3] or filtered back-projection algorithms[4][5]. Nevertheless the quality and the accuracy of the reconstruction need to be objectively evaluated. Different solutions can be considered such as the animal experimentation or the acquisition system modeling and the 3D physical or numerical phantom building.

We propose a simulation environment to evaluate the 3D reconstruction algorithm of the coronary arteries. It includes

- The construction of a 3D dynamic model of the coronary tree from dynamic volumes based on multi-slice computed tomography (MSCT) sequences (volumes reconstructed every 10% of the cardiac cycle). The result is a sequence of volumetric images representing the motion of the coronary arteries throughout the cardiac cycle;
- The modeling of the rotational angiography acquisition system to simulate different acquisition and gating strategies;
- The calculation of radiographic projections of the 3D model of coronary tree throughout several cardiac cycles.

This paper describes a preliminary version of this project: the generation of a geometric coronary tree model (section II), the modeling of the rotational angiography acquisition system and the calculation of projective data from the 3D coronary model (section III). Section IV concludes on the results and gives some perspectives.

II. CORONARY TREE MODEL GENERATION

A. Source Data

The model described in this paper is based on cardiac MSCT data. Dynamic volume sequences were acquired on a sub second spiral GE 64-slice CT scanner. Each sequence includes 10 volumes reconstructed at every 10% of the RR interval. The slice thickness is 0.625mm, the pixel size 0.488mm² and the size of the isotropic volumes is 512 × 512 × 332.

Although the 64-slice MSCT scanner gives rise to a higher temporal and spatial resolution for the visualization and the assessment of the coronary artery disease, some vascular segments remains not well displayed on several volumes of the sequence because of motion artifacts or severe wall calcification. This makes coronary arteries hard to extract
in these areas. And it is the reason why we focused for a preliminary stage on the generation of a static model from the volume reconstructed at the 80% of the R-R interval.

B. Centerline Extraction

We applied a tracing-based algorithm for the coronary artery central axis extraction[6]. The tracking algorithm relies on a local modeling of the vessel by a cylinder in a 3D homogeneous space. The parameters of this cylinder (location of the center of gravity, and diameter) are estimated using a 3D geometrical moment operator. A multiscale filter based on eigenvalue analysis of the Hessian matrix is then applied to highlight tubular structures and coping with varying widths. It is also used to locally determine the principal direction of the vessel. A detection of possible bifurcation is performed then in the estimated direction. If one bifurcation is detected, more than two seed points are extracted which are saved in a list to be further taken as initial points of the tracking process. The next point $P_{i+1}$ on the centerline is sought out in the estimated direction at the current point $P_i$ by using the moment operator.

C. Region Segmentation

We applied then the watershed algorithm around the extracted centerlines of the vessels to improve the accuracy of the vascular wall delineation. Watershed segmentation considers the regions to be extracted as catchment basins in topography. The idea of watershed is to fill up these basins with water by starting at local minima of gradient map. At the points where water coming from different basins meets, dams are built to constitute the boundaries of the basins or the watershed lines[7].

We performed region segmentation by using the following stages:

1) Define a volume of interest $B_i^{w}$ around the extracted centerlines. The bottom of the box is centered at each central axis point $P_i$. The width of this rectangular box is given by the vessel diameter $d_i$ estimated at the previous stage and the length $L_i$ by the span between two points $P_i$ and $P_{i+1}$. The orientation of the box $B_i^{w}$ was defined by the local orientation of the vessel $\vec{v}_i^{ex}(1)$;
2) Preprocess the volume with a gradient operator inside the $B_i^{w}$;
3) Apply 3D watershed algorithm by the Insight Toolkit(ITK) inside the box $B_i^{w}$;
4) Perform 3D region growing starting from the point $P_i$ to extract the vessel according to the label values of $P_i$.

Watershed algorithm in its original form produces an over-segmentation result of the image. To avoid this problem, watershed depth is introduced to overlook the small basins, which can be defined as the maximum depth of water a catchment basin can hold without flowing into any others[8]. Furthermore, another iterative strategy is performed to alleviate over-segmentation. Given the local vessel in $B_i^{w}$ is a frustum of a cone, its local volume $Vol_i$ can be calculated by local diameters $d_i$, $d_{i+1}$ and the distance $L_i$. Measured by the number of voxels, the calculated local volume $Vol_i$ can be compared with the volume of vessel region $Vol'_{i+1}$, segmented by watershed algorithm. The ratio $\gamma$ between these two volumes is thus defined as

$$\gamma = \frac{Vol'_{i+1}}{Vol_i}$$

Normally, $Vol'_{i+1}$ is larger than $Vol_i$, which makes $\gamma$ varying between 1 and 5. If it is less than 1, it suggests that over-segment may take place. The output scale of watershed algorithm will increase by adding a static value until the ratio $\gamma$ becomes a reasonable one.

Fig. 2 illustrates an extracted 3D coronary tree by 3D watershed method. The aorta was independently segmented with a classical region growing algorithm within a region of interest(ROI).

III. 3D ROTATIONAL ANGIOGRAPHY SIMULATION

A. System Geometry

The 3D-RA system is composed of an X-ray source and an image intensifier mounted on a motorized computer controlled arc known as the C-arm. The X-ray source is fixed with respect to the center of rotation (ISO Center). The distance source - image plane (SID) and the source-ISO Center distance (ISO) are two geometric parameters of the system. The entire assembly can rotate along an arc around a table, allowing multiple acquisitions from different angles(Fig. 3). Two angles, named primary angle $\theta$ and
secondary angle \( \varphi \) respectively, control the orientation of the C-arm. \( \varphi \) is kept constant over the acquisition time and varies from \(-30^\circ\) cranial to \(30^\circ\) caudal. \( \theta \) refers to rotation angle from RA060\(^{\circ}\) to LA060\(^{\circ}\) with angular increment (1.5\(^{\circ}\) - 5\(^{\circ}\)) applied during the rotation of the C-arm. Another parameter is available which allows selecting the rotation speed between \(30^\circ \text{s}^{-1}\) and \(40^\circ \text{s}^{-1}\).

The size of volume data and its spatial resolution, \( N_N^V \times N_N^V \times N_N^V \) and \((S_x^V, S_y^V, S_z^V)\), are provided by source data. Whereas, the size of projection image and its spatial resolution, \( N_N^I \times N_N^I \) and \((S_x^I, S_y^I)\), can be specified manually.

After analyzing the geometry of 3D-RA system, volume coordinates and projection image coordinates can be transformed into the system coordinates. Based on the right-hand system coordinates defined in Fig. 3, local volume coordinates \((x, y, z, 1)^T\) and projection image coordinates \((u, v, 0, 1)^T\) can be transformed to the corresponding system coordinates \((x_N^V, y_N^V, z_N^V, 1)^T\) and \((x_N^I, y_N^I, z_N^I, 1)^T\) by matrix \(R_V\) and \(R_I\) respectively. Because the axes of volume coordinates are parallel with the system axes, \(R_V\) can be simply defined by scaling and translating.

\[
R_V = \begin{bmatrix}
S_x^V & 0 & 0 & S_y^V \cdot V_{\text{Center}}^x \\
0 & S_y^V & 0 & S_y^V \cdot V_{\text{Center}}^y \\
0 & 0 & S_z^V & S_z^V \cdot V_{\text{Center}}^z \\
0 & 0 & 0 & 1
\end{bmatrix}
\]  \hspace{1cm} (2)

where \((V_{\text{Center}}^x, V_{\text{Center}}^y, V_{\text{Center}}^z)\) is the user-defined rotation center, which is coincided with the origin of system coordinates, i.e. ISO-Center. Composed of scaling, translation and rotation, \(R_I\) can be decomposed into matrix representation as: \(R_I = N_I \cdot M_I\). \(N_I\) represents the 3D rotation transform about an arbitrary unit vector \(\bar{u}\) which can be obtained by rotating \(+z\) axis by \(\varphi\) in \(yz\) plane. The detail about \(N_I\) calculation can be found in [9]. \(M_I\) relating to scaling and translation is defined as

\[
M_I = \begin{bmatrix}
S_x^I & 0 & 0 & -S_x^I(N_I - 1)/2 \\
0 & S_y^I & 0 & -S_y^I(N_I - 1)/2 \\
0 & 0 & S_z^I & (SID - ISO) \\
0 & 0 & 0 & 1
\end{bmatrix}
\]  \hspace{1cm} (3)

**B. Digitally Reconstructed Radiographs**

Based on volumetric data reconstructed by CT images, digitally reconstructed radiographs (DRR) can often be calculated for simulating x-ray projections. A ray-tracing based algorithm was adopted in [10][11] to calculate DRR based on the transmission principle of x-ray.

The concept of ray-tracing based algorithm is shown in Fig. 4. The ray coming from the x-ray source crosses the volume by intersecting a series of voxels before reaching the image plane. The coordinates of the intersection points \(I_i^\text{Center}\) and the distance \(d_{i}^{\text{inter}}\) between each two neighboring points are then calculated in system coordinates. The X-ray energy transmitted through a structure is given by:

\[
I = I_0 \sum_{i=1}^{M} e^{-\mu_i d_{i}^{\text{inter}}}
\]  \hspace{1cm} (4)

where \(I_0\) is the incident x-ray energy, \(M\) is the number of intersection points and \(\mu_i\) is linear attenuation coefficient for the CT voltage applied during the CT slices acquisition, which can be calculated by CT value of the voxel and the attenuation coefficient of water for the same CT voltage[11].

![Fig. 3: Geometry of 3D-RA system.](image)

![Fig. 4: The concept of DRRs simulation algorithm[11].](image)

During 3D-RA examination, a cardiac catheterization is performed to inject a contrast agent into the arteries in order to improve the contrast between the vessel and the other structures. Fig. 5(a) illustrates one frame of a real cardiac 3D-RA sequence with size \(1024 \times 1024\) and 0.13\(\text{mm}^2\) spatial resolution, in which left coronary artery (LCA) filled with contrast agent is shown with lower intensity than other structures.
coefficient $\mu^B_i$ is selected to generate the background. The attenuation coefficient can be defined by

\[
\begin{align*}
\mu^V_i &= a\mu_i + b, \text{ voxel belongs to the vessel region} \\
\mu^B_i &= a\mu_i, \text{ else}
\end{align*}
\]

where $a$ is specified less than 1 to decrease the $\mu_i$ and $b$ increases $\mu_i$ of the voxels in the vessel region. In particular, we defined $a, b$ equal to 0.1 and 0.2 respectively.

Fig. 5: (a): A frame of real 3D-RA sequence of LCA, (b)-(n): 3D-RA simulation of LCA in source data.

C. Simulation Results

In Fig. 5 (b) to (n), thirteen simulated projections of LCA in 3D artery model of Fig. 2 are picked out each 10° from RAO 60° to LAO 60°. Due to the high computation complexity, the size of volume was decreased to a lower one, $256 \times 256 \times 200$ with 0.66mm³ spatial resolution. According to the size of LCA, the simulated projection images were specified to be $330 \times 330$ pixels with 0.4nm² isotropic spatial resolution. In the simulated projections, the skeletons of coronary arteries were illustrated with similar image backgrounds. But the spatial resolution of volume with at most 0.33mm³ limited the resolution of simulated projections greatly. In addition, the manner of contrast agent injection makes the difference between simulated and real images. In the cardiac MSCT the blood with contrast agent flowed all over the body, while, in 3D-RA the contrast agent congregated in the interested branches at the imaging moment. As a result, the background of 3D-RA typically composed of vertebral column, costal bones and lung was blurred with cardiac cavities, aorta and pulmonary arteries in the simulated projections.

IV. Conclusion

In this paper, we have described a preliminary version of a simulated environment to evaluate the 3D reconstruction algorithm of moving structures in X-ray rotational angiography. This version performs a rotational acquisition of a static object. We are currently working on the construction of the 3D dynamic model of the coronary tree to be able to simulate the acquisition of a moving structure.

REFERENCES