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

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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 R-R interval. The slice thickness is 0.625mm, the pixel size 0.488mm² and the size of the isotropic volumes is $512 \times 512 \times 332$.

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...
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.
secondary angle $\varphi$ respectively, control the orientation of the C-arm. $\varphi$ is kept constant over the acquisition time and varies from $-30^\circ$ to $30^\circ$ caudal. $\theta$ refers to rotation angle from RAO$60^\circ$ to LAO$60^\circ$ 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 $30^\circ s^{-1}$ and $40^\circ s^{-1}$.

The size of volume data and its spatial resolution, $N_x \times N_y \times N_z$ 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, $N_x^I \times N_y^I$ 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 $(x^V, y^V, z^V, 1)^T$ and $(x^I, y^I, z^I, 1)^T$ 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.

$$R_V = \begin{bmatrix}
S_x^V & 0 & 0 & S_{x^V}^{V, \text{Center}} \\
0 & S_y^V & 0 & S_{y^V}^{V, \text{Center}} \\
0 & 0 & S_z^V & S_{z^V}^{V, \text{Center}} \\
0 & 0 & 0 & 1
\end{bmatrix}$$  \hspace{1cm} (2)

where $(V_x^{\text{Center}}, V_y^{\text{Center}}, V_z^{\text{Center}})$ 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: $R_I = N_I M_I$. $N_I$ represents the 3D rotation transform about an arbitrary unit vector $\vec{u}$ which can be give 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

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

B. Digitally Reconstructed Radiographs

Based on volumetric data reconstructed by CT images, digitally reconstructed radiographs (DRRs) can often be calculated for simulating x-ray projections. A ray-tracing based algorithm was adopted in [10][11] to calculate DRRs 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 $P^{\text{Inter}}_i$ and the distance $d_{i^{\text{Inter}}}$ between each two neighbored 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)
coefficient $\mu^B$ 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.

![A real 3D-RA frame](image)

**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 $10^\circ$ from RAO $60^\circ$ to LAO $60^\circ$. Due to the high computation complexity, the size of volume was decreased to a lower one, $256 \times 256 \times 200$ with 0.66mm$^3$ spatial resolution. According to the size of LCA, the simulated projection images were specified to be $330 \times 330$ pixels with 0.4mm$^2$ 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 0.33mm$^3$ 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**


