A reconstruction platform for coronary arteries, finite element mesh generation and patient specific simulations

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Abstract. The left coronary artery (LCA) is one of the most important sites of atherosclerotic plaque formation and its progression may lead to stroke. There are many systemic risk factors that are related to the appearance and development of atherosclerosis. It has been observed, however, that the lesions occur in specific regions of the arterial tree. Nowadays it is accepted that these regions are the ones with low abnormal values of Wall Shear Stress (WSS). The aim of this work is to generate a platform to obtain a three-dimensional (3D) reconstruction from an LCA, generate a finite element mesh and run a patient specific blood flow simulation in it. The 3D model of the LCA reconstructed was based on a computed tomography (CT) study from a healthy patient. The method used for this procedure consisted basically on three stages: (1) segmenting the LCA principal branches from TC images, (2) extracting several contours from the arteries so as to get the essence of the geometry, applying a self-developed software, and (3) using these outlines to generate a Non-uniform rotational basis spline (NURBS surface). A finite element unstructured volume mesh was generated and refined and a simulation was run on it using a Computational Fluid Dynamics (CFD) application. The values of pressure and velocity were analyzed and WSS was computed. As expected, regions of low WSS were found in the outer walls of the bifurcation and in the inner wall of curvatures.

1. Introduction

Atherosclerosis is currently one of the leading causes of mortality worldwide. It is a chronic, inflammatory, fibroproliferative disease primarily of large- and medium-sized conduit arteries (Chatzizisis et al., 2007). It is well known that there exist many risk factors that may contribute to the appearance of the atherosclerotic plaques, such as age, gender, hyperlipidemia, cigarette smoking, hypertension, diabetes mellitus, chronic infections, stress and genetic predisposition (Naghavi et al., 2003). Despite the systemic characteristics of these factors it was found that atherosclerotic lesions form in specific regions of the arterial tree, like in the vicinity of branch points, the outer wall of bifurcations and the inner wall of curvatures (Malek et. al., 2007). This seems to be related with the hemodynamic forces that depend on the individual geometry of each patient’s vessels. The most influencing of these forces is, by far, the tangential one that generates the blood flow in the inner arterial wall, and can be synthesized as the wall shear stress (WSS). Currently,
it is well accepted that the regions where plaques are more likely to emerge are those in which the flow bifurcates and there is low WSS (Davies et. al., 2009; Caro et. al., 2009; Steinman et. al., 2002). Moreover, for the particular case of the coronary arteries, recent studies showed that bifurcation angles in the in these arteries might influence atherosclerotic processes in the coronary Junctions (Chaichana et. al., 2010; Craiem et. al., 2009).

The fact that low WSS has an important role in the development and progression of atherosclerosis lies in the endothelial response: it attenuates nitric oxide (NO)-dependent atheroprotection; it promotes low-density lipoprotein cholesterol (LDL) uptake, synthesis, and permeability; it promotes inflammation and oxidative stress; it promotes vascular smooth muscle cell (VSMC) migration, differentiation, and proliferation; it promotes ECM degradation and attenuates its synthesis in vascular wall and plaque fibrous cap; it has a potential role in plaque neovascularization and calcification; it increases plaque thrombogenicity (Chatzizisis et al., 2007).

Computational Fluid Dynamic (CFD) modelling using the Finite Element Method (FEM) has been proved to be a practical and reliable tool for studying time-varying, three-dimensional (3D) blood flow patterns in complex arterial geometries. It has been shown to provide prognostically relevant hemodynamic data based on 3D reconstructions of vessels from medical images (Steinman et. al.,2002). It is imperative to use patient-specific reconstructions of arteries and make them as reliable as possible. The FEM is a numerical technique for solving partial differential equations (PDE’s). Its first essential characteristic is that the continuum field is subdivided into cells, called elements, which form a grid. (Wendt et. al., 2009)

In this paper we present a method to reconstruct human coronary arteries from Computed Tomography (CT) images using a self-developed software. We use then the results of this reconstruction platform to generate a finite element volume mesh from a Left Coronary Artery (LCA) to run a FEM simulation of the blood flow in it.

2. Methods

2.1. 3D surface reconstruction

A critical issue to obtain reliable simulations is the reconstruction procedure used to model the vessels. We propose a method that consists basically on three steps: (1) segmentation of the principal LCA branches from CT images; (2) extraction of several contours from the arteries in order to characterize the geometry; (3) use of these outlines to generate a Non-uniform rotational basis spline (NURBS) surface.

2.1.1. CT segmentation. In the present work we used a multislice CT scan performed to an asymptomatic and healthy patient in the Hospital Universitario, Fundación Favaloro, Buenos Aires, Argentina. The scan was done using a 64-row MSCT scanner (Aquilion 64, Toshiba, Japan). Images were acquired during diastole by means of ECG-gating using a 400–960 ms time window in order to reduce motion facts. Transaxial images were reconstructed with 0.415 mm slice thickness and 0.3 mm increments.

All digital imaging processes were done using a custom software designed in the Favaloro University. A detailed explanation of this processing is explained elsewhere (Craiem et. al., 2011). We imported with the mentioned software 40 consecutive images (512 by 512 pixels, 256 gray levels) so as to reconstruct the whole LCA, beginning in the ostium, including the left main artery (LM), the left circumflex (LCX) and the left anterior descending (LAD) arteries until an important secondary branch in any of them appeared (Figure 2) (Salvucci et. al., 2010).

The LCA extraction proceeding is based on two main stages: Heart isolation and coronary artery segmentation.

The first phase, where the heart is isolated from the images, starts with the user selecting a volume of interest (VOI) where the whole heart must be included. The heart isolation consists in an iterative
process of 5 steps and concludes in the clustering of every pixel between Heart-Region and Not-Heart-Region, being the values of the last group set to the minimum value of the whole volume in Hounsfield Units (HU).

The second stage (the segmentation of the coronary artery), involves 6 steps which include a 3D median filter, a 3D morphological opening filter, digital substraction, contrast increase, a windowing process and a volumetric region-growing algorithm. If at this point some elements as coronary veins, ventricle blood pools regions near papillary muscles or spurious noise connected to the LCA remain visible, the software provides a manual correction 3D erasing tool to remove them.

Figure 2: (A) Aorta and LCA anatomy. (B) Example of a CT slice image showing the coronary ostium, the left main artery (LM), the left anterior descending (LAD) and the left circumflex (LCX) branches.

2.1.2. Contour extraction. The idea is to obtain the contour of the LCA, with planes orthogonal to the midline (skeleton) of the vessels, in order to get the artery 3D geometry. For this purpose we used a self-developed software implemented in Matlab®. This routine has 5 algorithms: skeletonization, branch division, surface triangulation, contours computation and IGES exportation.

The skeletonization method is a sequential 3D thinning algorithm adapted from the one developed by Palágyi K. (Palágyi et. al., 2001). It is basically an iterative process that iteratively eliminates certain points form the solid object until a single pixel skeleton is found. It analyzes the 26-neighbourhood of every voxel and determines thereby if it belongs or not to the surface of the remaining solid. Each iteration in the algorithm itself is based on 6 subiterations that allow classifying the border points. Figure 3 (A) shows a solid reconstruction of the LCA used in this investigation and (B) its skeleton.

The branch division algorithm implemented in this platform begins by automatically classifying points of the skeleton in bifurcation-points, end-points and middle-points (Figure 3 (C)). This is done by counting the points in the immediate vicinity of each skeleton point. The user is then asked to select the beginning-point and the end-point of each branch. Then, the software compares every point of the solid LCA geometry and associates each point to the nearest branch as shown in Figure 3 (D).

For accomplishing the surface triangulation, the software implements the Marching Cubes algorithm. This procedure is used to render isosurfaces in volumetric data (Bourke et. al. 1994). A result of applying this algorithm is shown in Figure 3 (E).

The contours computation is done with the intersection between planes orthogonal to the vessel centerline and the marching cubes volume. The centerline is estimated using a spline curve. The user manually sets the spline order and the number of points to be analyzed in each segment. A representative artery is shown in Figure 3 (F)

Finally, the IGES exportation step is simply an algorithm that uses the information from the contours to generate an .iges file that can be recognized by any CAD software as spline curves in 3D.
2.1.3. NURBS surface generation. This phase was done using a custom developed CAD software. First of all the .iges file with the splines of the arterial contours are imported (Figure 4 (A)). The two branches of the bifurcation have to be lofted separately, sharing the curves from the main branch (LM) as shown in Figure 4 (B) and (C). Finally, these two surfaces are joined into the final surface, including the inlet and outlet patches (Figure 4 (D) and Figure 4 (E)).
Figure 4: (A) Spline Curves representing the contours of the LCA. (B) Lofting of one side of the arterial bifurcation. (C) Lofting of the other half. (D) Trimming the surfaces leaving an empty space between both halves. (E) Blending the bifurcation and generating a closed polysurface.

2.2. Mesh generation and refinement
For generating the mesh we used Netgen® software designed by Schöberl J.. The mesh size and granularity were set so as to obtained a useful mesh for our investigation. In particular, the mesh size was set smaller in the proximity of pronounced curves. The mesh was then exported to enGrid® software. Here we created a prismatic element boundary layer in the vicinity of the arterial wall and we divided it in order to get a final outcome with a very thin boundary layer. This is the key point for an accurate calculation of the WSS. The obtained mesh was exported to the OpenFoam format.

2.3. Computational Fluid Dynamics
Blood flow through the LCA was simulated using the finite element code OpenFoam (OpenCFD Ltd.). A SIMPLE solver was used to solve the governing three-dimensional Navier-Stokes equations to obtain the velocity vector \( \vec{v} \) and pressure \( p \) in every point of the mesh. Flow was assumed to be steady and laminar. Blood was considered as a homogeneous Newtonian fluid with a dynamic viscosity of \( \mu = 0.0035 \text{ Pa}\times\text{s} \) and incompressible with a density of \( \rho = 1050 \text{ kg/m}^3 \). The vessel wall was modeled as rigid (Salvucci et. al., 2010).

The simulation was done setting \( p \) in the LCX outlet patch to zero and imposing \( \vec{v} \) in the inlet and LAD outlet patches with a surface normal planar velocity profile. Flow splitting was calculated according to Murray’s cubic law (Soulis et. al., 2006). Free traction condition was set to the LCX outlet patch and \( \vec{v} \) was set to zero at the vessel walls according to the non-slip condition.

Once the simulation was performed we computed WSS from the gradients of the \( \vec{v} \) fields near the wall:

\[
WSS = \frac{\mu}{\rho} \left| \frac{d\vec{v}}{dn} \right|
\]

where \( n \) is the vector normal to the wall in each point.

3. Results and Discussion
In Figure 5 we show the mesh obtained following the procedure described in the previous section. Particularly, in (A) we present the mesh as it was created and refined with Netgen® and in (B), the final result, after generating and dividing the prismatic boundary layer in the vicinity of the arterial wall with enGrid®. This was the mesh used for running the simulation.

Figure 5: (A) Mesh obtained with Netgen®, (B) Final mesh used for the simulation with a thin prismatic boundary layer generated with enGrid®.
The results of the simulation run with OpenFoam are exhibited in Figure 6. The first panel (A) shows the pressure values computed in this LCA reconstruction. OpenFoam uses a “cinematic pressure”, this is the pressure divided by the fluid density. These relative values are all calculated taking into account that pressure was set to zero in the LCX output as the boundary conditions were defined. In panel (B) a velocity vector graph is shown. Finally, in panel (C), the WSS distribution is presented. As expected, regions of very low WSS were found in the outer walls of the bifurcation (Caro et al., 2009; Soulis et al., 2006). Furthermore, other zones of low WSS were detected in the three branches in the inner wall of curvatures as reported by others (Malek et al., 2007).

![Figure 6: Blood Flow simulation results: (A) Pressure, (B) Velocity and (C) WSS.](image)

One of the main limitations of this work is that it considers the flow as stationary and, in consequence, it doesn’t include the effects caused by the oscillations. An accurate determination of the WSS is obtained when combining the pulsatile (unsteady) nature of the arterial blood flow with the complex geometric configuration of the coronaries (Chatzizisis et al., 2007). Furthermore, this investigation doesn’t take into account the interaction of the blood flow with the vessel’s wall, which plays an important role in the appearance and development of the plaques (Malek et al., 2007). Besides, here were consider only a small segment of the coronary arterial tree, not looking at the effects that may bring the surroundings.

In spite of these limitations, we show that coronary arteries can be reconstructed in vivo from MSCT images and flow profiles can be simulated with realistic results. User intervention was minimized and computational cost was reduced. The time spent for the segmentation and 3D surface reconstruction was about 30 minutes, and for configuring the simulation, 5 minutes. The time that takes the simulation to run is highly variable and depends on the particular features of the computer, the characteristics of the mesh and the CFD configuration. In this case, using an average laptop, it took about 3 hours. All these results indicate that we walk towards a true 3D individual characterization of the coronary tree geometry and WSS profile that will be applied to better understand atherosclerosis progression.

4. Conclusion

In the present work we present a platform and a methodology to perform a 3D surface reconstruction from CT images of LCA and generate a finite element volume mesh of it. The whole process itself has many automatic steps but it requires also a lot of user interaction. This is what guaranties a reliable geometry.

We studied a specific case, performed the corresponding reconstruction and used the resulting mesh to perform a FEM simulation. The pressure, velocity and WSS obtained correspond,
qualitatively, to the expected according to the bibliography. Moreover, regions of low WSS where found in the predicted zones.

Concerning the limitations of this work, further investigations should be done considering a pulsatile flow, it’s interaction with the vessel’s wall and taking into account a bigger portion of the coronary arterial tree, not just the first bifurcation.

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References
