Impact of Surface Mesh Simplification on Hemodynamic Simulations

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Abstract. In this paper we present an approach to quantify the effect of triangular surface simplification on hemodynamic modeling. This work is organized in three parts. First, we briefly present the basic concepts inherent in hemodynamic modeling. Then, we describe the approach of triangular surface simplification. The simplified triangular surfaces are used as input as well as a parabolic inlet velocity profile to the hemodynamic modeling procedure. As a validation, the third part will be focused on two examples carotid blood flow characterization for a healthy patient and one with stenosis in the carotid artery. We quantified the differences between the generated models by evaluating the eucledean similarity, rate of change and RMSE between the measured and modeled velocity obtained with different mesh densities. We conclude with the important effect of re-meshing on the modeled results and we propose a minimum threshold of facets which ensures a good simulated velocity profile with a minimum error rate.

Keywords: hemodynamic modeling, triangular surface simplification, surface mesh simplification

1 INTRODUCTION

Triangular meshes are a very common representation for 3D surfaces. They enable the creation of highly detailed models and the application of various processing operations such as surface reconstruction, matching, shape modeling and analysis, denoising [1], mesh simplification [2], surface-modeling to measure for example the quality of the approximation of smooth surfaces from coarse meshes [3], to quantify local deformations in the LV of the heart [4] or to visualize vascular structures by exploiting the local curvature information of a given surface [5]. The present work is interested in investigating the effect of geometry on the quality of hemodynamic simulations in the vascular structure.

Although it is well established that that blood velocity may be affected by the vessel's resistance, pressure, viscosity and diameter, the present work is interested in investigating the effect of geometry (triangular meshes) on the quality of hemodynamic simulations in the vascular structure. We focused on triangular surface simplification of a given triangle mesh surface embedded in R^3. As a first step, we started by implementing a process of triangular mesh simplification. Afterwards, we used the different generated geometries as input of a hemodynamic simulator. An evaluation of the simplification effect on the quality of the simulation will be carried out.

The paper is organized as follows. The basic concepts inherent in hemodynamic modeling are depicted in the second section. In section three, the approach of triangular surface simplification is presented. The results obtained from real cases of the carotid artery of a healthy subject and a patient with a stenosis are outlined in section 4. An assessment of the effect of mesh simplification on the hemodynamic modeling is performed by computing some indicators (section 5). The results were then discussed and a conclusion summarizes the present work and presents our future perspective.

2 HEMODYNAMIC MODELING

The numerical method used in blood flow modeling is described by the following incompressible Navier-Stokes equations [6,7] :

$$\nabla . v = 0 \tag{1}$$

$$\rho \left(\frac{\partial v}{\partial t} + v . \nabla v\right) = -\nabla p + \nabla . \tau \tag{2}$$

Where v is the velocity, ρ the density, t the time, p the pressure and τ the stress tensor given by the following equation :

$$\tau = \mu\dot{\gamma} \quad (3)$$

Where μ is the dynamic viscosity, and $\dot{\gamma}$ the strain rate tensor.
For Newtonian fluid, the dynamic viscosity is a constant.
For Non Newtonian fluid, based on Herschel–Bulkley model, viscosity is defined as:

$$\mu = k\dot{\gamma}^{n-1} + \frac{\tau_0}{\dot{\gamma}} \quad (4)$$

Where k is the consistency index, n is the power-law index and τ_0 is the yield stress threshold.

To perform this modeling, we needed to use an initial velocity at the input. Different works have focused on the impact of velocity profiles on the hemodynamic simulations [8]. In this investigation, we applied a parabolic inlet velocity (Fig. 1) which was set via the ANSYS User Define Function (UDF) with the following expression:

$$v = h\left[1 - \left(\frac{(x-x_c)^2+(y-y_c)^2}{R^2}\right)\right] \quad (5)$$

Where *h* is the parabolic altitude corresponding to the value of artery center velocity given by US Doppler recording in CCA. x_c and y_c are the CCA artery center coordinates and *R* is the CCA radius (fig. 1).

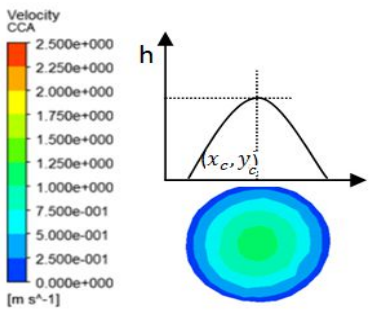


Fig. 1. Example of numerical simulations of parabolic inlet velocity.

3 TRIANGULAR MESH SIMPLIFICATION

Surface re-meshing aims to generate new triangle meshes of the original surface by re-sampling the input mesh removing (mesh simplification) or adding (mesh refinement) vertices. In this work a mesh simplification method is used.

Triangle mesh simplification aims to generate simplified models approximating the original ones. Several algorithms have been proposed in this context and can be mainly divided into three classes: (i) vertex decimation methods based on vertex classification and re-triangulation schemes, and preserving the topology of the input model [9], (ii) vertex clustering [10] and iterative edge collapse [11].

4 INFLUENCE OF MESH DENSITY ON HEMODYNAMICS

4.1 Data Acquisition

Two types of data were acquired from two patients:
a) A 3D Computed Tomography Angiography (CTA) image for the vessel modeling. The 3D-CTA acquisition contains 41 slices and 132 slices for the first and the second patient respectively. Each slice contains 512 x 512 pixels. The spatial resolution is 0.782 x 0.782 x 1.0 mm for the first patient and 0.506 x 0.506 x 0.5 mm for the second one. Patient 1 was healthy and patient 2 was diagnosed with stenosis at ICA.
b) A Pulsed Doppler Ultrasound (US) acquisition will be used to set the boundary conditions of hemodynamic modeling and evaluate blood flow simulations. The profiles were taken at the center of the artery (Fig.2). The central value (Mid-systole) for patient 1 was 112 cm/s and 86 cm/s for patient 2.

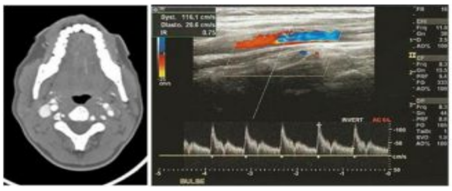
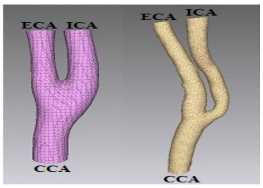


Fig. 2. Acquisitions: CTA slice (left) and Doppler Ultrasound image (right) of the carotid artery.

4.2 Iso-surface reconstruction and remeshing process

To construct the patient's 3D vascular model, we started by segmenting the 2D CTA slices, then extracted the triangulated mesh representing the surface of the 3D structure using the Marching cubes algorithm [12]. As a result, we obtained the 3D carotid construction of the two patients. The mesh relative to the first patient is composed of 4380 faces and 2156 vertices. The mesh referring to the second one (P2) is composed of 20158 faces and 10081vertices (fig. 3).



(a)-Patient 1 (b)-Patient 2

Fig. 3. Generated Triangular mesh for the right carotid bifurcation of the two patients: (a) Patient 1- 4308 faces and 2156 vertices, (b) Patient 2- 20158 faces and 10081 vertices.

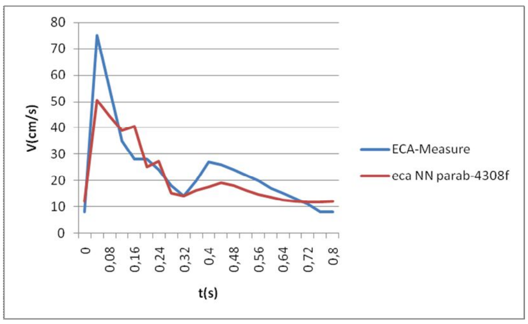
4.3 Blood flow simulation according to surface mesh

We aimed at modeling blood flow and estimating the effect of surface mesh simplification on the quality of hemodynamic simulations of blood, particularly, in the carotid artery. The latter is composed of the Common Carotid Artery (CCA), which is derived from the aorta and bifurcates to the External Carotid Artery (ECA) and the Internal Carotid Artery (ICA).

Our process uses mainly two inlets:
1-Geometric data (triangular meshes) (fig. 3).
2-Velocity profile of the CCA (fig. 2), which will be subsequently useful to ensure that the modeled velocity profile is as close as possible to the real one.

4.3.1. Hemodynamic simulation

Three localizations were studied: CCA, ECA and ICA. CCA velocity profile was used as an inlet velocity profile toward a hemodynamic behavior as close as possible to the real one. As a blood flow model, we compared the actual profiles to the simulated ones in the case of Non Newtonian fluid. We chose a parabolic inlet flow since it is efficient for such carotid bifurcation modeling [13] (fig. 4, 5). We used ANSYS-FLUENT to solve the blood flow modeling equations by the finite volume method (FVM). For Non Newtonian fluid k=0.01 Kg/m-s, n=0.68 and τ_0=0.4 Pa. Density ρ was set to 1060 Kg/m3. For each mesh $M_{i=1..4}$.



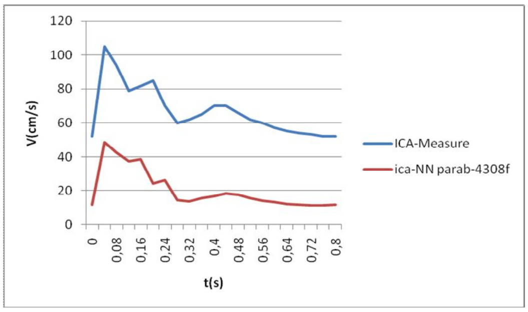


Fig. 4. Simulated and measured velocity profiles in ECA (top) and ICA (bottom) for patient 1.

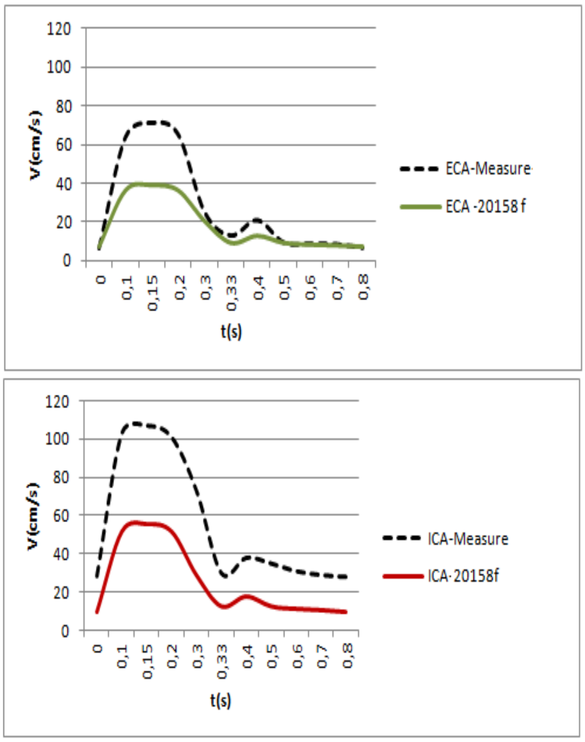
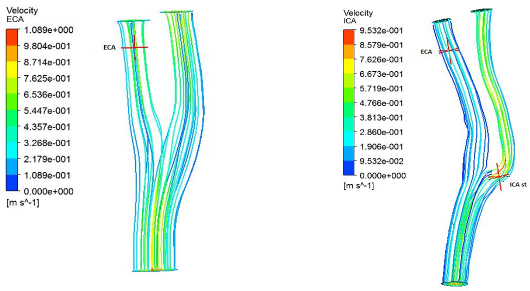


Fig. 5. Simulated and measured velocity profiles in ECA (top) and ICA (bottom) for patient 2.



(a) (b)

Fig. 6. Simulated velocities and localisation of mesearement points for patient 1 (a) and patient (2).

Figure 6 shows the results of the simulated velocity for both patients and the approximate positions of the measurement points performed with Doppler Ultrasound. Subsequently, all the simulated velocity profiles will be generated at the two points at the external carotodian arteries ECA.

4.3.2. Similarity measures

In order to quantify the differences between the measures and the simulated velocity profiles, three indicators were calculated.

a) The similarity (Sim) between two series X (x_1, x_2,..., x_n) and Y (y_1, y_2, ..., y_n) can be calculated as follows [13]:

$$\mathrm{Sim}(x), y = 1/\sqrt{\sum_{i=1}^{n}(xi - yi)^2} \tag{6}$$

b) The rate of change (RC) between two profiles is given by :

$$RC = \frac{norm\,(Xmeasure - Ysimulated)}{norm\,(Xmeasure)} \tag{7}$$

c) The Root Means Square Error (RMSE) is calculated as follows :

$$RMSE = \frac{1}{N}\sum_{i=1}^{N}\frac{\sqrt{(X_i^{measure} - Y_i^{Simulated})^2}}{max\,(X^{measure})} \tag{8}$$

4.3.3. Conclusion on hemodynamic simulation in relation to geometric input

Table 1 gives the results of three computed indicators between mesured and simulated velocity in ECA and ICA for the two patients.

Table 1. Comparison between measured velocity and simulated velocity for patient-1 and patient-2.

Artery	RC %	Sim	RMSE %
ECA-P1	26.9759	0.0286	2.2200
ICA-P1	68.6604	0.0046	9.7603
ECA-P2	42.5697	0.0191	6.6437
ICA-P2	52.3061	0.0091	9.3799

It can be concluded that the model produces better results for the ECA simulated velocity, without modifying any hemodynamic parameters and using a parabolic input. In the next section, we will detail the effect of remeshing on the ECA blood flow simulations.

5 INFLUENCE OF MESH SIMPLIFICATION ON HEMODYNAMICS

This part deals with the impact of mesh simplification in order to deduce which facet minimal rate per region unit will produce the model closest to the reference. The idea was to choose the maximum-density meshing that will produce a result closest to the model.

We applied our mesh simplification procedure in simulations of blood flow in a carotid artery of a specific patient. To assess the influence of the surface remeshing on hemodynamics, we proposed a method consisting of three steps(i) iso-surface reconstruction, (ii) Mesh simplification and (iii) Velocity profile modeling and hemodynamic simulation.

We started from the initial mesh of the carotid composed of 4380 faces and 2156 vertices for patient1 (MI-P1) (fig.7) and 20158 faces and 10081 vertices for patient2 (MI-P2) (fig. 8). We generated four simplified meshes for patient1 and patient 2 (Table 2). The 3D surface of the carotid artery for patient 1 is 12.6 (cm2) and is 18.9 (cm2) for patient 2.

Table 2. The generated simplified triangular meshes for the patient 1 and patient 2.

Triangular mesh	Mesh P1-I	Mesh P1-1	Mesh P1-2	Mesh P1-3	Mesh P1-4
Faces	4308	1722	860	172	86
Vertices	2156	863	432	88	45
Surface (cm2)	12.57				12.61
Triangular Mesh	Mesh P2-I	Mesh P2-1	Mesh P2-2	Mesh P2-3	Mesh P2-4
Faces	20158	8060	4030	860	200
Vertices	10081	4032	2017	432	102
Surface (cm2)	18.91				18.88

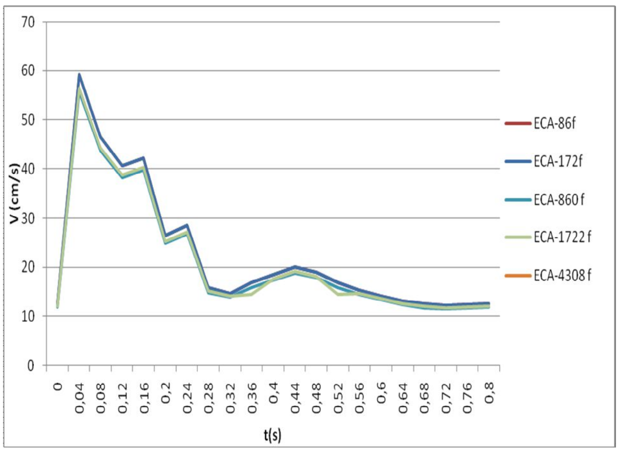


Fig. 7 . Patient 1 : Simulated blood flow models for the ECA parabolic velocity for initial mesh (4308f), four simplified meshes 1722f, 860f, 172f, and 86f.

Table 3. Comparison between ECA velocity profile for the initial mesh (4308f) and the simplified meshes, 1722f, 860f, 172f and 86f for Patient 1.

ECA Simulated	RC %	Sim	1-Sim	RMSE %
ECA-1722f	5.7829	0.1561	0.8439	0.6045
ECA-860f	4.8869	0.1848	0.8152	0.5108
ECA-172f	8.9853	0.1005	0.8995	0.9392
ECA-86f	29.0408	0.0311	0.9689	3.0356

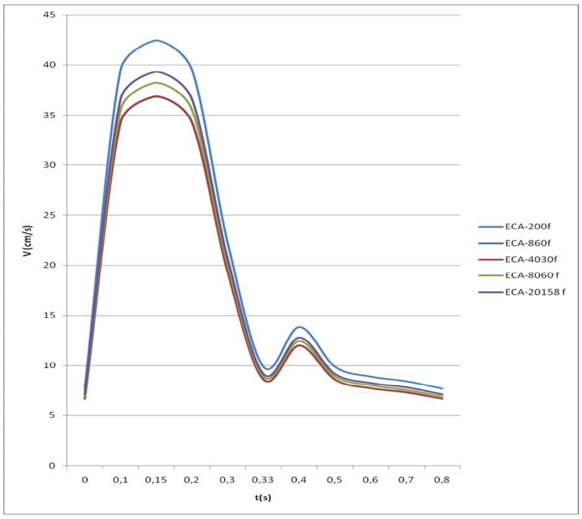
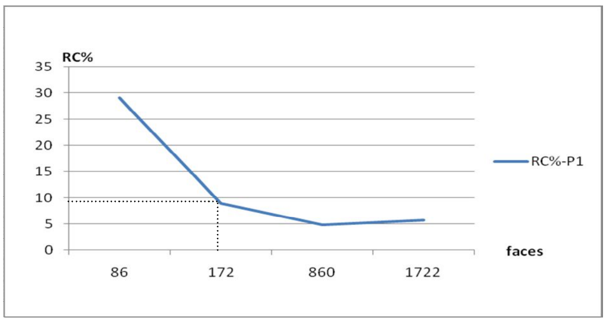


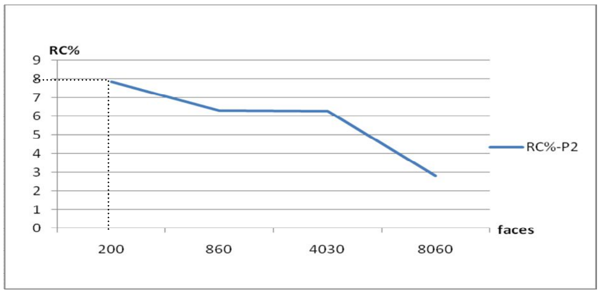
Fig. 8 . Patient 2 : Simulated blood flow models for the ECA parabolic velocity for initial mesh (20158f), four simplified meshes 8060f, 4030f, 860f and 200f.

Table 4. Comparison between ECA velocity profile for the initial mesh (20158f) and the simplified meshes, 8060f, 4030f, 860f, and 200f for Patient 2.

ECA Simulated	RC %	Sim	1-Sim	RMSE %
ECA- 8060f	2.8242	0.4907	0,5093	0.4711
ECA-4030f	6.2657	0.2212	0,7788	1.0451
ECA-860f	6.3099	0.2196	0,7804	1.0525
ECA-200f	7.8731	0.1760	0,824	1.3133



(a)



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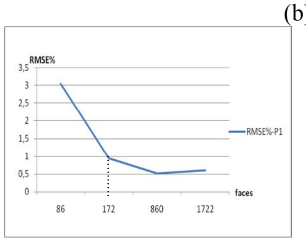
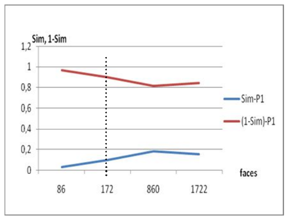
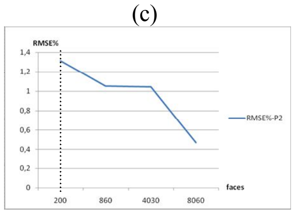


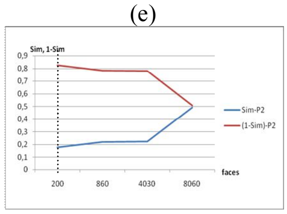
(c)



(e)



(d)



(f)

Fig. 9. Variations of the three indicators in function of the number of facets for the two patients, RC - patient1 (a) and RC - patient2 (b), RMSE - patient1 (c), RMSE - patient2 (d), S and 1-Sim-patient1 (e), S and 1-Sim-patient2 (f).

Figure 9 shows the variations of the three indicators in function of the number of facets for the two patients, RC-patient1 (fig.9.(a)), RC-patient2 (fig.9.(b)), RMSE-patient1(fig 9.(c)), RMSE-patient2 (fig 9.(d)), Sim and 1-Sim-patient1 (fig.9.(e)), Sim and 1-Sim-patient2 (fig.9.(f)). The reading of these different curves enabled us to conclude that the minimum number of facets which allows a correct simulated velocity profile is 170 for the first patient and 200 for the second one both at RC=8%. The minimum threshold of facets, which ensures a good simulated velocity profile with a minimum error rate, is defined as the number of facets by unit of surface. This threshold is 170/11.82=14.38 for patient 1 and 200/18.9=10.58 for patient 2. The adopted threshold was 15 facets per cm^2.

6 DISCUSSION

In this paper we ensured a hemodynamic simulation in specific conditions. We considered blood as Non Newtonian fluid, fixed all hemodynamic parameters and used a parabolic velocity input. The only modified input was density of surface mesh.

In this framework, we generated some examples of simplified meshes. We chose three indicators (Rate Change RC, Similarity and RMSE) in order to quantify the differences between different velocity profiles. We concluded that simplification of surface triangular mesh has an important effect on the hemodynamic modeling results.

This work enabled us to propose a minimum threshold of facets which ensures a good simulated velocity profile with a minimum error rate. This threshold is defined as the number of facets by unit of surface, and the proposed value is 15 facets by cm^2. Using a number of facets bigger than this threshold does not significantly improve the quality of the velocity profile obtained but will result in a significant increase in computation time. In the near future, we want to (i) analyze the relation of this threshold with the other variables considered as constant in this work, and (ii) locally remesh the surface to optimize velocity profile simulation.

To improve the quality of our results and to cover the majority of cases we needed to study many real cases and to propose a flexible arterial model for validation. The principal limitation of this work is the reference velocity obtained by Doppler Ultrasound measurement. We are now working with another type of data characterized by a satisfying spatial resolution and a great number of velocity measure points obtained from 3D phase contrast MRI.

7 CONCLUSION

The effect of surface simplification on the quality of hemodynamic simulations was investigated in the present study. We fixed all hemodynamic parameters and we used only a parabolic velocity input. Three indicators were computed in order to quantify the effect of the mesh surface variation on velocity profile. To validate our findings, we performed a hemodynamic analysis on the healthy patient and the one with stenosis.

The first results enabled us to propose a minimal threshold in number of facets per unit surface, which ensures an acceptable simulated velocity profile. To improve and confirm our results, we are now working to enrich our patient database, integrate other geometric and hemodynamic parameters and above all to resort to more accurate measurements of velocity profiles such as MRI.

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