Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Fluid/Structure Interaction applied to the simulation of Abdominal Aortic Aneurysms

Jean-Luc Pélerin*‡, Carine Kulik†, Cemil Goksu† Jean-Louis Coatrieux†, Michel Rochette‡,

*École Centrale Nantes, 1 rue de la Noë BP 92101, 44321 Nantes Cedex 3, France
†Laboratoire du Traitement du Signal et de l'Image, INSERM, Université Rennes I
Campus de Beaulieu, 35042 Rennes Cedex, France
‡ANSYS France SAS, 11 avenue Albert Einstein, 69100 Villeurbanne, France

***Abstract*— Aneurysms are a local dilatation of a vessel wall of at least twice the normal diameter (commonly accepted definition). They are asymptomatic and rupture is often lethal. Thus, prediction of rupture is an important stake. Aiming at a diagnosis tool relying on patient specific data and general physiological values, we created a virtual aneurysm model based on real imaging data. Fluid/structure interaction (FSI) simulations were made to compute the displacement and stress for the wall. For the fluid, the only in vivo measures used were for the inlet velocity. The mandatory output boundary condition has been implemented with the first order Windkessel model equations. Structure has been much more complicated to handle because of the association of a realistic geometry (no symmetry) and a full fluid/structure interaction approach. We used surface elements to stabilize the structure and to model surrounding organs. Validation parameters are the displacement, the Von Mises stress and the pressure profile at the outlet. The main difference with other studies such as [1], [2], [3] relies on the association of all these elements in order to prepare industrial applications as the main goal of this study was to build an automated tool easy to use by people who are not experts in numerical simulation.**

***Index Terms*— Abdominal Aortic Aneurysm, Fluid Structure Interaction, Patient specific simulation**

I. INTRODUCTION

ANEURYSMS are not dangerous until they rupture. In such a case the death rate is between 50 and 70% even if operated [4]. That is why it is important to develop tools for rupture prediction and to help clinicians to decide whether it is necessary to operate or not.

Moreover as surgery becomes less and less invasive, surgeons need more visualization tools. This study is a first step towards integration of Computational Fluid Dynamics (CFD) and Fluid/Structure Interaction (FSI) to existing pre-operative assistance software developed at the Laboratoire du Traitement du Signal et de l'Image (LTSI). The two ways extended FSI capabilities of ANSYS/CFX added in version 10 and all the software suite has been used to build a virtual model of an Abdominal Aortic Aneurysm. We paid much attention to the realism of the geometrical model as it is intended to be used in a patient specific oriented project.

AAAs are very good candidates for this kind of simulation test case as it is known that the diameter, the commonly used criteria to evaluate rupture risks, is less accurate than vessel wall stress [2]. Thus, this kind of simulation can provide information of prime interest to clinicians that they cannot directly infer from image volume.

II. THE MODEL

A. Structure

1) Geometry: Geometry has been extracted semi-automatically from multi detector computer tomography using previously written software from the LTSI and based on the gradient method [5]. This processing stage provides data that are organized into a set of planes each containing 48 points belonging to a particular surface (artery's inner an outer wall limit, thrombus limit). Each set is homotopic to a cylinder (each representing a possible path through the arterial tree). Because we decided to include the iliac bifurcation, this processing had to be able to remove redundant data.

Then, the resulting point sets had to be corrected for three reasons:

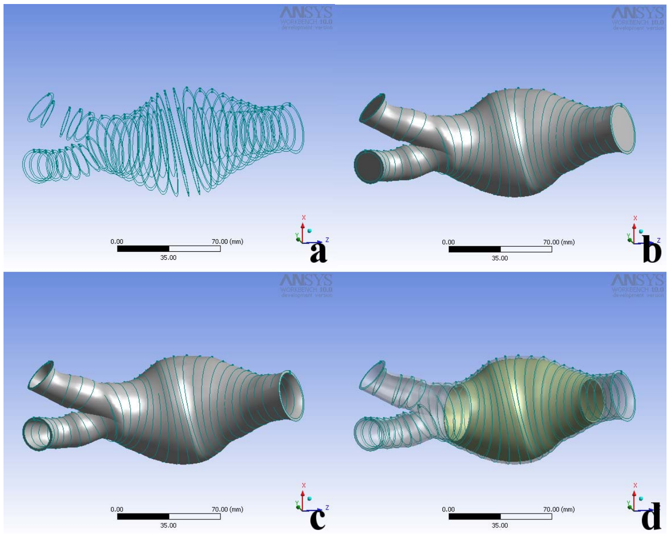
- First, because of the errors committed by the segmentation algorithm (most of the errors are located at the interface between two structures not being detected by the gradient algorithm)
- then, because flow would suffer simulation artifacts (such as local vortex) if the wall is not as smooth as it really is
- and also because CAD software has difficulties to deal with local surface singularities

A few errors were first corrected manually, then a simple relaxed method converging towards a circle was applied to the points sets, thus, making it possible to choose how smooth the model is.

Finally, this processing had to output data in a format suitable to ANSYS Workbench which was used for geometry reconstruction (see figure 1), meshing and data setup (boundary conditions) steps. Using Workbench Constructive Solid Geometry(CSG) capabilities, three domains were created: the arterial wall, the thrombus sac and the blood vein.

2) Mechanical properties: Mechanical properties are summarized in table I and are taken from [6] and [7].

1-4244-0033-3/06/$20.00 ©2006 IEEE.



a) Geometry lies on splines. b) A filled volume is created
c) A boolean operation is used to build the fluid path
d) Thrombus is separated from the wall

Fig. 1. Steps of the geometry reconstruction

Property	symbol	value	unit
Density	ρ	1100	$\mathrm{kg.m^{-3}}$
Young modulus	E	1.2	MPa
Poisson ratio	ν	0.49	–
Mesh Foundation Stiffness coefficient	E'	150000	–

TABLE I
MECHANICAL PROPERTIES

3) Boundary conditions: Departing from other studies that adopted an FSI strategy, our geometric model does not assume any property such as symmetry or privileged direction. Thus, it appeared to be very difficult to find boundary conditions that would satisfy the following closely interacting requirements:

- The structure must be enough constrained so that the solver can converge and avoid global displacement.
- Boundary conditions at the extremities must allow wave propagation (a “no displacement” boundary condition is excluded)
- surrounding tissues that sustain the artery must be simulated to avoid stretching of the artery
- No constraint discontinuity on the wall to avoid unrealistic high stresses
- Other nodes should be free

In regard to the extremities, it is important that the structure is not constrained axially to eliminate wave reflections. That is why nodes of the extremities were just obliged to stay in their initial plane and able to move normally to the axis.

But it is not enough as the structure is underconstrained. The systolic peak would stretch the artery unrealistically. The proposed solution was to cover the external arterial wall with surface elements (called “Elastic Foundation Stiffness” in ANSYS) that behave as springs uniformly spread on the elements. These elements simulate the surrounding tissues. (See figure 2).

Property	symbol	value	unit
Density	ρ	1050	$\mathrm{kg.m^{-3}}$
Compressibility	β	36.0	$\mathrm{m^2.s^{-2}}$

TABLE II
BLOOD PROPERTIES

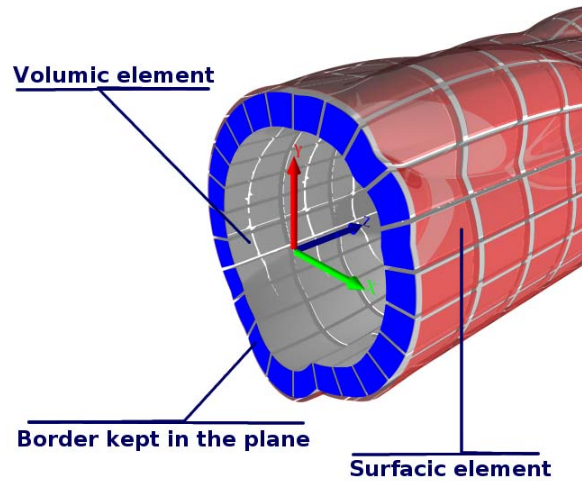


Fig. 2. Illustration of mesh and boundary conditions for each extremity

B. Fluid

1) Blood rheology: The blood has been considered as a Newtonian fluid. Blood properties used are summed up in table II. A simple $k - \varepsilon$ turbulence model has been used. Blood has also been considered has slightly compressible for numerical reasons (for an incompressible fluid, a small wall displacement would lead to a large change in pressure)

2) Boundary conditions:

a) Inlet: The velocity at the inlet was imposed uniformly all over the inlet using transient values taken from in vivo measures, these values were found in [7]. A quadratic interpolation was used. Figure 3 presents the profile used.

b) Wall: A no slip boundary condition has been applied on the wall.

c) Outlet: For the outlet, a Windkessel model has been implemented using CFX ability to accept external code. The Windkessel model consists in considering the downstream part of the arterial system as an elastic chamber. A first order model was adopted.

The Windkessel model can be compared to an RC circuit with a first order differential equation :

$$Q = \frac{1}{R}P + C\frac{\partial P}{\partial t} \tag{1}$$

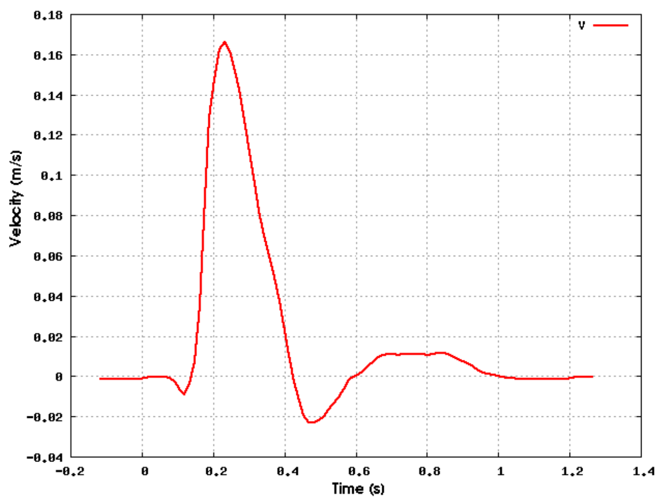


Fig. 3. Blood velocity (m/s) in the aorta (inlet boundary condition)

It was discretized using finite differences :

$$Q^{n+1} = \frac{1}{R}P^{n+1} - C\frac{P^{n+1} - P^{n}}{\mathrm{d}t} \quad (2)$$

The convergence of this equation is ensured by the fact it is included inside CFX coefficient loop.

C. Coupling

Coupling implementation is a "strong" coupling : there is an inner convergence loop for the coupling inside the timestep loop. This loop, called the stagger loop, ensures that each field exchanged has converged (in our case, Pressure and Displacement).

III. Results

For a preliminary validation of our simulation system with the Windkessel model, the maximum displacement and the maximum stress in the arterial wall have been used.

A. Output pressure

The output pressure has been compared to the one given in the same study we took the velocity profile from ([7]). The comparison is depicted on figure (4). Because the model is very simple (only first order), we took into account only the global shape and the scale. We found back the main characteristics : the systolic peak, a fast decrease to half the maximum and a slow decrease to pressure at rest. The slow decrease is very important because it is the main difference with the velocity profile and also because it is linked to the elasticity of blood vessels and stiffness of surface elements.

B. Wall Displacement

1) Global displacement: It is important to ensure that the mean displacement of the whole model is not too important because a large global displacement would mean that the structure is underconstrained. It could have been a problem as we used only surface elements to maintain the structure.

We observed that the displacement of the centroid has a maximum of 0.2 mm. It is perfectly acceptable considering the use of surface elements.

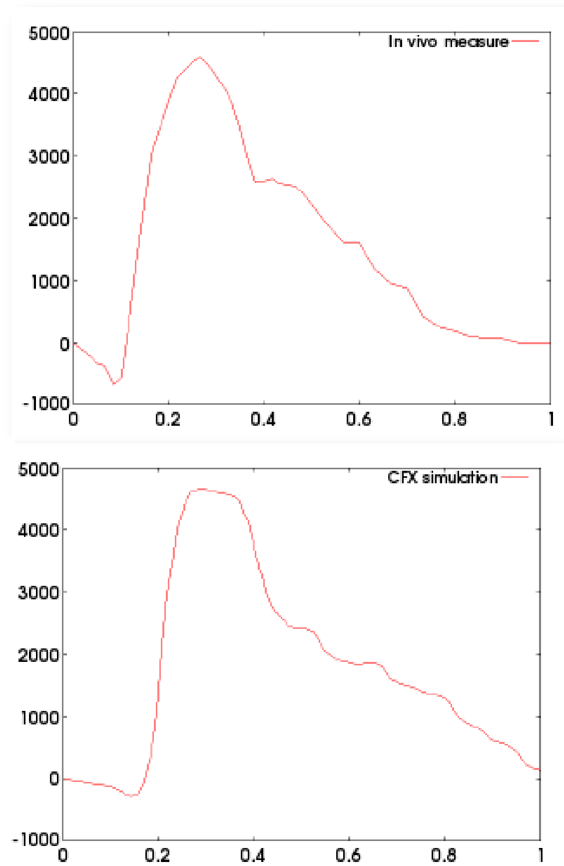


Fig. 4. Comparison of output pressure (in mmHg) between in vivo measures (top) and computed values (bottom)

2) Local displacement: Wall displacement is shown on figure 6. It has been compared to [7]. Results are similar:

- Same displacement repartition (Maximum displacement at the inlet)
- Maximum Value : 2.2 mm against 2.3 mm

The difference of position can be explained by the fact we did not imposed a null displacement at the inlet.

C. Maximum vessel wall stress

The validation the wall stress parameter is much more difficult as we lacked data to compare our results to : in vivo measures are inaccessible, existing values are also computed and do not refer to the same geometry.

However, the maximum Von Mises computed was 47.890 N/cm^{-2}. And the study [8] gave values higher than 44 N/cm^{2} for aneurysms with the same diameter (6.6 cm) and the same shape.

IV. Conclusion - Future Work

This study shows that it is possible to associate a full fluid-structure interaction simulation, a patient specific geometry, boundary conditions that are either general or that do not require heavy in vivo measures. These conditions are prior to a clinical/industrial application. An original structure boundary condition has been introduced with "Elastic Foundation

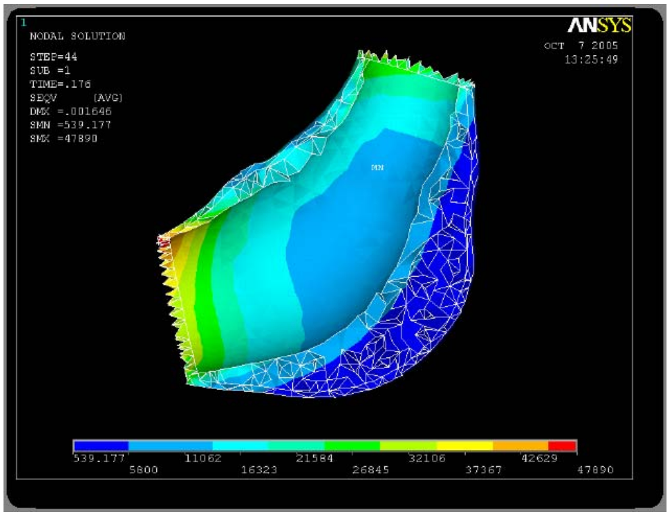


Fig. 5. Von Mises stress (Part of the vessel wall and thrombus)

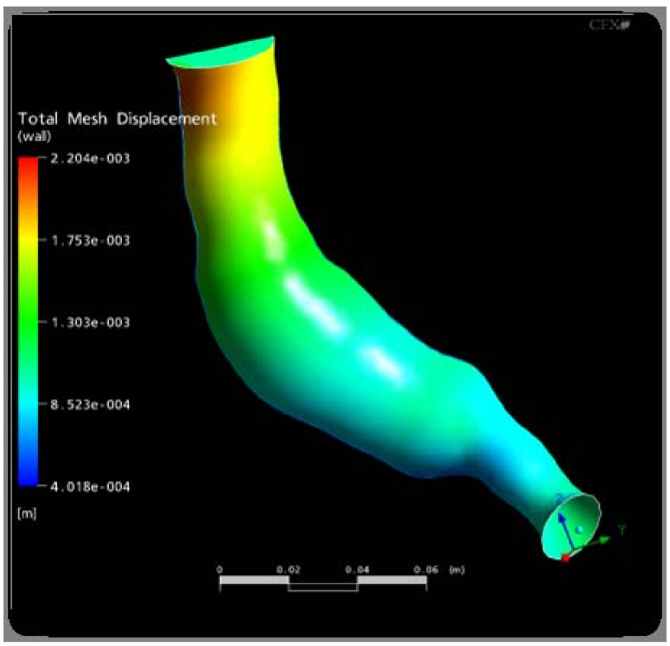


Fig. 6. Wall displacement (Fluid domain only)

Stiffness elements" that facilitates the use of non symmetrical geometry.

However, some improvements still have to be carried out on the following points :

- The oscillations observed on the output pressure profile has to be eliminated through a better choice for the Elastic Foundation Stiffness coefficient (a viscoelastic model should probably give better results).
- A better model should be used for the vessel wall as it is known to be non linear and inhomogeneous. We have already written a tool able to set local properties derived directly from scanner images.
- Segmentation results can be improved to alleviate the preprocessing stage.
- The Windkessel model gives good results, but more sophisticated (and precise) models exist.
- Many parameters were considered as constant whereas they should be modifiable. For example : arterial pressure, arterial wall state, hematocrit, etc.

V. Aknowledgements

Many people were associated directly or indirectly to this study. We would like to thank each of them.

Sadek Cherhabili who has assisted us on the CFX and Coupling data setup procedure and shared his experience and knowledge.

The whole team at ANSYS UK which gave us a warm welcome and dedicated time.

Clement Kleinstreuer who accepted to answer our questions.

Pascal Haigron who followed our work and made important suggestions.

And the ANSYS France technical support team which helped us for data setup and software use.

References

[1] C. Scotti, A. Shkolnik, S. Muluk, and E. Finol, "Fluid-structure interaction in abdominal aortic aneurysms: effects of asymmetry and wall thickness," *BioMedical Engineering OnLine*, vol. 4, no. 1, p. 64, 2005. [Online]. Available: http://www.biomedical-engineering-online.com/content/4/1/64

[2] M. F. Fillinger, S. P. Marra, M. L. Raghavan, and F. E. Kennedy, "Prediction of rupture risk in abdominal aortic aneurysm during observation: Wall stress versus diameter," *Journal of Vascular Surgery, Volume 37, Number 4*, 2003.

[3] Y. Papaharilaou, J. A. Ekaterinaris, E. Manousaki, and A. Katsamouris, "Stress analysis in abdominal aortic aneurysms applying flow induced wall pressure," *5th GRACM International Congress on Computational Mechanics*.

[4] P. Kergosien, "Traitement endovasculaire des anévrismes de l'aorte abdominale sous rénale," Ph.D. dissertation, Faculté de Médecine de Rennes, 2004.

[5] P. Haigron, M. Bellemare, O. Acosta, C. Goksu, C. Kulik, K. Rioual, and A. Lucas, "Depth-map-based scene analysis for active navigation in virtual endoscopy," *Transactions on Medical Imaging*, vol. 11, no. 23, pp. 1380–1390, 2004.

[6] F. Duck, *Physical properties of tissues: a comprehensive reference book*. Academic Press, London, 1991.

[7] Z. Li and C. Kleinstreuer, "Blod flow and structure interactions in a stented abdominal aortic aneurysm model," *Medical Engineering & Physics*, 2005.

[8] M. Rhagavan, D. Vorp, M. Federle, M. Makaroun, and M. Webster, "Wall stress distribution on three-dimensionnaly reconstructed models of human abdominal aortic aneurysm," *Journal of Vascular Surgery*, vol. 31, no. 4, pp. 760–769, 2000.