Patient-specific finite element simulation of the insertion of guidewires during an EVAR procedure: towards clinically relevant indicators

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Résumé :

Ce travail présente une méthode de simulation par éléments finis explicite pour le calcul des déformations d'une structure vasculaire aorto-iliaque induites par l'insertion de guides endovasculaires de type « extra-stiff » pour le traitement des anévrismes abdominaux. Le modèle mécanique prend en compte le comportement non-linéaire de la paroi vasculaire, l'effet de précontrainte induit par la pression artérielle et le support mécanique dû aux organes et structures environnants. Les résultats de simulation sont confrontés à des données d'imagerie 3D acquises au cours de la procédure chirurgicale sur 24 patients. Ces résultats sont ensuite utilisés afin de déduire des quantités utiles d'un point de vue clinique, comme le raccourcissement de segments artériels, la longueur des prothèses à déployer, le déplacement de points anatomiques importants.

Abstract :

This work presents an explicit FE method to compute the deformations of an aorto-iliac structure induced by the insertion of endovascular extra-stiff guidewires for the treatment of abdominal aneurysms. The mechanical model takes into account the nonlinear behavior of the arterial wall, the prestressing effect induced by the blood pressure and the mechanical support of the surrounding organs and structures. The simulation results are evaluated against 3D imaging data acquired during the surgical procedure on 24 patients. Then the results can be used to derive clinically relevant quantities, like arterial segment shrinkage, length of the stent-graft to be deployed, displacement of important anatomical points.

Mots clefs : Abdominal Aortic Aneurysm, Endovascular repair

1 Introduction

EndoVascular Aneurysm Repair (EVAR) is a commonly used mini-invasive technique which has gained increasing popularity over the last 10 years. It relies on the exclusion of the aneurysm sac by means of the femoral introduction of one or more stent grafts and their deployment inside the aneurysm. During the intervention, insertion of several endovascular tools is required to offer support and stability to the stent-graft delivery system. Among them, insertion of extra-stiff guidewires often leads to the straightening of vascular structure. During this process, the vascular structure undergoes major deformations [1]. Usually, these have no consequence on the smooth progress of the procedure. However, in some instances, the deformations caused by the insertion of stiff guidewires can have major consequences. Straightening tortuous arteries may lead to shrinkage of the arterial segments compared to the preoperative anatomy, thus calling into question the initial sizing of the stent grafts. It can also induces major movements of anatomical markers such as the ostia of the digestive arteries, leading, in the most critical cases, to a risk of covering a secondary artery [2]. Conversely, in some cases, the tools are incapable of straightening excessively calcified and tortuous arteries, thus precluding the delivery of the stent graft, which usually leads to an amendment of the procedure or to its cancellation [3].

Arterial deformations caused by the endovascular equipment during surgery depend on multiple factors, such as the morphology of the arteries, the state and degree of calcification of the arterial wall, and also the types of devices used. Today, their prediction relies mainly on the surgeon’s experience. Numerical simulation appears to be an appropriate tool to anticipate the complications mentioned above [4][5]. Its use in the preoperative phase could give the practitioner more objective and more useful indicators when planning the procedure: guiding the surgical act and making it safer using such an approach would potentially reduce the risks of intraoperative and postoperative complications.

Here, we present the development of a method for the mechanical simulation of the vascular structure deformations due to the insertion of extra-stiff guidewires during an EVAR procedure using an explicit finite-element software. The simulation results are evaluated against 3D imaging data acquired during the surgical procedure on 24 patients. Once the predictive capability of the simulation is attested, we can derive clinically relevant quantities, like arterial segment shrinkage, length of the stent-graft to be deployed, guidewire curvature, and displacement of important anatomical points like ostia of secondary arteries.

2 Methodology

2.1 Biomechanical model

- **Vascular structure modeling**

The vascular geometry represented in the simulations corresponds to an aorto-iliac structure including the abdominal aorta and the common and external iliac arteries as far as the femoral bifurcations. The models are meshed with triangular shell elements. The behavior of the arterial wall is modeled using the polynomial, nonlinear and isotropic hyperelastic potential defined by Equation (1):

$$W = C_{10}(I_1 - 3) + C_{20}(I_1 - 3)^2 \quad (1)$$

With C_{10} = 0.005 MPa and C_{20} = 0.2 MPa. Elements corresponding to calcification plaques are linear elastic properties (Young’s modulus E = 40 MPa, Poisson’s ratio υ = 0.4). Intraluminal thrombus was not represented in the models.

- **Loading and boundary conditions**

The effect of surrounding organs and structures is modeled by a visco-elastic support on the entire surface of the vascular mesh. The resulting nodal load is defined by equation (2):

$$\vec{f}_{ext} = \vec{t}_{ext}.dS = (-k\vec{u} - c\vec{v} + p\vec{n}).dS \qquad (2)$$

Where $\vec{u}$ is the displacement of the considered mesh node, $\vec{v}$ the velocity of the node, $\vec{n}$ the vector normal to the surface at this node, k the surfacic stiffness coefficient of the support, c the surfacic viscosity coefficient of the support, p the external pressure, $\vec{f}_{ext}$ the resultant external force vector and dS the elementary surface. The elastic stiffness k represents the support from the surrounding organs and tissues. Its value is locally defined on the surface of the arterial mesh based on the local distance from the arterial wall to the spine and the position along arterial segments. An elastic stiffness k_{iliac} is also added at the beginning of the internal iliac arteries in order to represent the limited mobility of these points. The relatively stationary proximal and distal extremities are modeled by a zero-displacement condition over the three outer borders of the mesh.

The prestressing effect of blood pressure is also taken into account. We use an iterative method similar to the one described by Bols and al. [6] to determine a new geometry, which corresponds to a stress-free state often called “zero-pressure geometry”. Then, at the start of the simulation, this geometry is subjected to internal pressure, which enables one to recover the reference geometry as observed on the CT-scan images, except that it now incorporates the prestress due to blood pressure. Then we process with the simulation of guidewire insertion.

The guidewires represented in the simulation are Lunderquist “extra-stiff” (Cook), they are modeled using 189 4mm-long two-noded beam elements with linear elastic properties (Young’s modulus E=180 GPa , Poisson’s ratio υ= 0.3) The tools are pushed at the distal end of the external iliac arteries by prescribing a velocity at their lower end until they are fully inserted in the vascular structure.

2.2 Results validation and derivation of clinically relevant data

As very little literature data is available concerning tethering of aorto-iliac vessels, values of the parameters defining the elastic support were first calibrated through sensitivity analysis on one patient-specific case [7], the results are then blindly evaluated on 24 other patient-cases to assess the method robustness and predictive capability. For each patient, either a 3D rotational acquisition or multi-incidence 2D acquisitions are done under fluoroscopy imaging, which allows reconstructing the 3D position of the real intraoperative guidewire. After registration of the intraoperative data into the simulation coordinate system, the positions of the simulated guidewire and the real guidewire can be compared. The metric used to measure the error between the simulated and the real guidewire position is the Modified Hausdorff Distance (MHD) [8].

Once simulation accuracy is asserted through guidewire position error measurement, several clinically relevant quantities can be derived from the simulation results:

- A measure of the guidewire curvature once inserted into the vascular structure is estimated by the computation of its flexural energy normalized by its length.
- The change in common iliac length is measured as the length difference between the preoperative centerline and the centerline of the artery deformed by the guidewire.
- The deployment length is defined by the length of the guidewire portion comprised between the renal arteries ostia and the internal iliac artery ostium and corresponds to the length on which the stent-grafts are going to be deployed.
- Finally, the displacement of left and right renal arteries ostia is measured between the preoperative and the deformed configurations.

Among the 23 patient-cases, for 7 of them intraoperative images acquisition was done in 3D with contrast agent injection allowing for visualization of the real 3D configuration of deformed arteries. For theses 7 cases, the 4 quantities defined above were thus compared to measurements made on real deformed intraoperative arterial configuration. For the other cases, 3D intraoperative acquisition only provides the guidewires position.

3 Results and Conclusions

Table 1 reports simulation results for 24 patient-cases, values of MHD and guidewire curvatures show that, whereas the calibration was done on only one patient-specific case, the simulation results are very good for 13 cases with an error lower than 5 mm. For 4 cases the error is greater than 10 mm. Guidewire curvature is very well predicted for most of the cases and can be a relevant indicator to tell the surgeon if there is a risk to face difficulties to deliver the stent-graft to its deployment site, as reported in [3]. Table 2 displays the computed values of common iliac change of length, deployment length, and renal arteries displacement for the 7 contrast-enhanced cases. Results show that in the two most severe cases common iliac change of length is well predicted. However there is a high uncertainty in the measure of real intraoperative length due to segmentation, registration and point picking accuracy so the results are not easy to interpret. The same applies to renal artery displacement that is too low in comparison to measure uncertainty to allow drawing any conclusions. On the other hand, deployment length is well predicted for all the cases and could be a very good indicator for surgeon to anticipate the length of the stent-graft to be implanted.

Finally, the work reported here shows that numerical simulation can predict some rather large variations in the vascular geometry due to tools insertion. Along with the final deformed configuration of the vascular structure that can be used to improve intraoperative visualization and guidance, the preoperative computation of some quantities described here could help surgeons in planning their interventions, by providing more objective arguments for sizing and choice of the stent-grafts and by anticipating possible intraoperative complications.

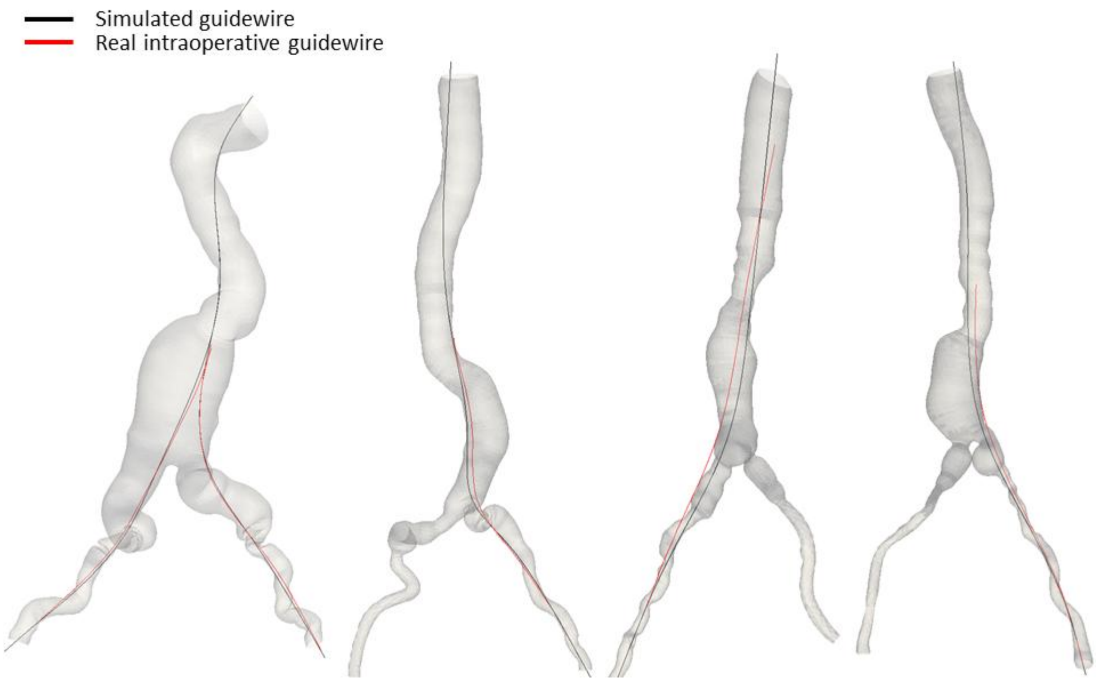

Figure 1: Illustration of real and simulated guidewires position comparison on 4 patient-cases

Patient id	MHD (mm)	Guidewire curvature from the simulation (flexural energy by unit length in J/mm)	Measured guidewire curvature (intraoperative) (flexural energy by unit length in J/mm)	Guidewire curvature relative error
1	3.44	5.7	4.8	18.8%
2	3.65	14.9	13.6	9.6%
3	5.95	3.1	2.7	14.8%
4	5.13	7.3	8.8	-17%
5	3.29	3.3	2.5	32%
6	2.3	5.5	4.8	14.6%
7	1.48	5.6	5.3	5.7%
8	2.94	6.3	6.4	-1.6%
9	5.64	16.5	16.9	-2.4%
10	5.21	7	4.9	42.9%
11	3.48	1.5	4.4	-65.9%
12	3.25	7.5	6	25%
13	10	5.3	9.6	-44.8%
14	3.14	7.6	7.7	7.5%
15	5.66	1.9	2.4	-20.8%
16	6.66	3.8	4	-5%
17	11.6	9.6	5.4	77.8%
18	3.6	9.1	8.9	2.2%
19	3.15	7.4	6.9	7.2%
20	10.7	6.8	10	-32%
21	6.27	6.1	12.5	-51.2%
22	3.96	5.7	10.1	-43.6%
23	10	4.8	4.7	2.1%
24	2.53	0.8	2.1	-61.9%
mean	**5.13**			**25.3% (absolute values)**

Table 1: Modified Hausdorff Distance (MHD) between real and simulated guidewire and curvatures computation for the 24 patient-cases

Patient id	Common iliac length change (simulation)	Common iliac length change (intraoperative)	Deployment length (simulation)	Deployment length (intraoperative)	Renal ostia displacement (left/right) (simulation)		Renal ostia displacement (left/right) (intraoperative)	
1	-12.9	-12.8	144	142	0.2	-0.3	8.6	9.1
2	-13.3	-12.7	148	155	0.7	0.8	5.8	7.3
3	-6.4	0.1	156	162	-0.6	-0.3	5.8	6.2
4	2	-5	160	156	0.6	0.8	-8	-5
5	-4	1.7	144	147	-1	-0.6	2.6	2.4
6	1.1	-1	168	168	0.1	-0.6	3.2	3.1
7	-7.1	-0.1	132	142	0.3	0.5	1,1	2.8

Table 2: Clinically relevant quantities derived from simulation results for the 7 contrast-enhanced patient-cases

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