



HAL
open science

Finite Element Analysis of the Mechanical Performances of 8 Marketed Aortic Stent-Grafts

Nicolas Demanget, Ambroise Duprey, Pierre Badel, Laurent Orgéas, Stéphane Avril, Christian Geindreau, Jean Noël Albertini, Jean-Pierre Favre

► **To cite this version:**

Nicolas Demanget, Ambroise Duprey, Pierre Badel, Laurent Orgéas, Stéphane Avril, et al.. Finite Element Analysis of the Mechanical Performances of 8 Marketed Aortic Stent-Grafts. *Journal of endovascular therapy*, 2013, 20 (4), pp. 523-535. 10.1583/12-4063.1 . hal-00880940

HAL Id: hal-00880940

<https://hal.science/hal-00880940>

Submitted on 7 Nov 2013

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

TITLE PAGE

Authors

Nicolas Demanget, MSc^{1,2}, Ambroise Duprey, MD², Pierre Badel, PhD¹, Laurent Orgéas, PhD³, Stéphane Avril, PhD¹, Christian Geindreau, PhD³, , Jean-Noël Albertini, MD PhD², Jean-Pierre Favre, MD PhD²

¹ *Ecole Nationale Supérieure des Mines*

CIS-EMSE, CNRS UMR 5146

158 cours Fauriel

Saint-Etienne F-42023, France.

² *CHU Hôpital Nord*

Department of Vascular Surgery

Saint-Etienne F-42055, France.

³ *Laboratoire Sols-Solides-Structures-Risques (3SR Lab)*

CNRS UMR 5521, Université Joseph Fourier - Grenoble 1, Grenoble-INP

Domaine universitaire, BP53

Grenoble 38041 Cedex 9, France.

Title

Finite element analysis of the mechanical performances of eight marketed aortic stent-grafts

Acknowledgements

The authors gratefully acknowledge the Région Rhône-Alpes for its financial support.

Corresponding author

Nicolas Demanget

158 cours Fauriel F-42023 Saint-Etienne cedex 2

Tel: +33(0)477499772

E-mail address: nicolas.demanget@gmail.com

Conflict of interest

Dr. Jean Noël Albertini is a proctor for Cook Medical.

ABSTRACT AND KEY WORDS

Purpose

The mechanical behaviour of aortic stent-grafts (SGs) plays an important role in SG durability and thus the success of endovascular surgery. As this behaviour is still not very well understood, the aim of this study was to assess numerically the flexibility and the mechanical stresses undergone by stents and fabric of current manufactured SGs.

Methods

Eight marketed SG limbs (*Aorfix*[®], *Anaconda*[®], *Endurant*[®], *Excluder*[®], *Talent*[®], *Zenith Flex*[®], *Zenith LP*[®], *Zenith Spiral-Z*[®]) were modelled using finite element analysis. A numerical benchmark combining bending up to 180° and pressurisation at 150 mmHg of the SGs was performed. SG flexibility was assessed by the calculation of the luminal reduction rate (LR_{max}). Maximal stresses in stents (σ_S^{max}) and maximal strains in fabric (ε_{LG} and ε_{CG}) were also assessed.

Results

LR_{max} at 90° was less than 20% except for the Talent SG. LR_{max} at 180° was higher for Z-stented SGs (range 39 - 78%) than spiral or circular-stented SGs (range 14 - 26%). At 180°, σ_S^{max} was higher for Z-stented SGs (range 370 – 622 MPa) than spiral or circular-stented SGs (range 177 – 368 MPa). At 90° and 180°, strains in fabric were low and did not differ significantly between PET SGs (range 0.5 - 7%). ePTFE fabric of the Excluder SG underwent higher strains (range 11 - 18 %).

Conclusions

Stent design strongly influences mechanical performances of aortic stent-grafts. Spiral and circular stents provide greater flexibility as well as lower stress values than Z-stents, and thus better durability.

Key words

- Aortic stent-graft
- Finite element analysis
- Aortic aneurysm
- Mechanical behaviour.

INTRODUCTION

Endovascular repair (EVAR) is a widely used technique to treat abdominal aortic aneurysms (AAAs). However, stent-graft (SG) durability remains the principal issue of EVAR. Endoleaks^{1,2}, stenosis or thrombosis of the SG^{3,4}, SG components failure⁵⁻⁷ may require secondary interventions in up to 22% of cases at 5 years⁸. In tortuous AAAs, a lack of SG flexibility has been associated with the above-mentioned complications^{9,10}. However, few objective data are available regarding flexibility and mechanical stresses/strains in components of current marketed SGs.

Numerical modelling by Finite Element Analysis (FEA) can be used to compare different SGs by assessing and analyzing SG deformation but also stresses and strains undergone by the device components. Some studies have proven the feasibility of this approach by using homogeneous models (a single equivalent material used for both stent and graft) as a first approach¹¹⁻²⁰. These models were employed to investigate the interaction between SG and blood flow using Computational Fluid Dynamics (CFD) and Fluid Structure Interaction (FSI) techniques. Particularly, displacement forces acting on the SG and stresses within the AAA wall after SG placement were estimated. However, homogeneous SG models did not take into account the mechanical complexity of the device and therefore may yield to erroneous results. Stents and graft were not differentiated and, consequently, the interaction between the two materials was not taken into account. Therefore, these models were not suitable to compare SG flexibility and mechanical stresses in components.

More recently, more realistic multi-material SG models were proposed²¹⁻²⁴. To date, very few studies have used this approach. Kleinstreuer et al. (2008)²¹ were the first to propose such numerical models. In their work, a tubular diamond-shaped SG was subjected to a cyclic

pressure loading. Stresses in stent were assessed for two types of Nitinol and stresses in the graft were studied for PET and ePTFE. However, the stent-graft model did not reproduce specifically any marketed SG at that time. Furthermore, fabric modelling was simplified using isotropic materials. In 2012, deployment of a bifurcated Talent SG (Medtronic) within an aneurysm was simulated²⁴. Geometrical numerical results were validated by the means of X-ray microtomography of the SG deployed within a silicone aneurysm phantom. However, in this study, numerical results were exploited geometrically but not from the mechanical point of view.

Our group published recently a FEA study of the bending behaviour of Zenith Flex (Cook Medical) and Aorfix (Lombard Medical) iliac limbs with proper mechanical properties for SG components materials²². Spiral-stented limbs (Aorfix) were more flexible than the Z-stented limbs (Zenith Flex). Moreover, stresses in stents and strain in fabric were lower for spiral-stented limbs. This approach was rigorously validated experimentally from qualitative and quantitative points of view by using X-ray microtomography²³.

The aim of this study was therefore to extend the analysis of flexibility and stresses to all currently marketed SG.

METHODS

Stent-grafts

Eight marketed SG limbs were modelled (see figure 1):

- SGs with several Z-shaped stents:
 - Zenith Flex (Ze-SG) (*Cook Medical Europe, Bjaeverskov, Denmark*),
 - Talent (Ta-SG) (*Medtronic, Minneapolis, USA*),
 - Endurant (En-SG) (*Medtronic*),
 - Zenith Low Profile (Zlp-SG) (*Cook Medical Europe*).
- SG with single spiral stent:
 - Aorfix (Ao-SG) (*Lombard Medical, Didcot, United Kingdom*).
- SGs with single Z-spiral stent:
 - Excluder (Ex-SG) (*Gore, Flagstaff, USA*),
 - Zenith Spiral Z (Zs-SG) (*Cook Medical Europe*).
- SGs with several circular stents:
 - Anaconda (An-SG) (*Vascutek, Inchinnan, United Kingdom*).

SG iliac limbs were chosen because they are usually subjected to important deformations during and after their deployment within the iliac arteries. Dimensions of SG limbs and their components (stents and graft) were measured on samples obtained from manufacturer documentation. SG limb features have been reported in table 1.

Stent-graft modelling

SG modelling has been presented in detail and validated experimentally in previous papers^{22,23}. Thus, in the present paper, only essential details are recalled and some new features have been introduced. FEA software was Abaqus 6.8/Explicit[®] (Simulia).

Geometry and mesh

Geometrical and computational features of fabric and stents are reported in table 2. In order to ensure consistent comparison between devices, graft lengths (L_G) were slightly modified from the original samples and ranged between 88 mm and 108.1 mm. Cylindrical grafts were modelled and meshed with Abaqus 6.8.2. Geometries and meshes of idealized 3D-stents were generated with Matlab R2011a (MathWorks).

Stents

Each stent centreline was approximated by parametric equations. A homemade Matlab routine generated discretization of stent centrelines through these equations. Then triangular mesh of each stent cross-section and global mesh of the 3D-stent using 6-node linear triangular prism elements were generated (figure 2).

Fabric and sutures

Cylindrical grafts were meshed with orthotropic elastic linear shell elements (thickness = 0.08 mm) through the Abaqus mesh algorithm. Excluder graft thickness was lower than other graft thicknesses and was equal to 0.04 mm.

Sutures securing stents and graft together were not modelled in order to reduce computational complexity. Instead, a kinematic bonding between stent and graft outer surfaces was prescribed (“tie constraint” in Abaqus) so that both entities could not slide or

separate during simulations. Besides, self-penetration of SG components was also avoided through a self-contact algorithm.

Material properties

Stents

The constitutive behaviour of the 316L stainless steel Ze-SG stents was characterized mechanically through a homemade tensile test. An elastoplastic model with isotropic strain hardening was considered. Corresponding properties of this model are listed in table 3.

Other stents were made up of Nitinol (NiTi), whose superelastic and isotropic behaviour was modelled using the Abaqus subroutine originally proposed by Auricchio and Taylor (1997)²⁵. The same mechanical properties were used for all Nitinol wires since the elastic limit of the Nitinol was never reached during our simulations (the strain remained small). Besides, the Nitinol model used in our simulation is not symmetrical in traction/compression (transformation in compression: 585 MPa). Material properties listed in table 4 were taken from Kleinstreuer et al. (2008)²¹.

Fabric

In this study, two types of fabric were considered and meshed with shell elements:

- (i) Ex-SG graft consisted of ePTFE fabric whose properties, listed in table 5, were taken from Catanese et al. (1999)²⁶ and Kleinstreuer et al. (2008)²¹. This fabric was considered as isotropic elastic linear. It was assumed that the helical strip of the Excluder was made of the same ePTFE than the graft of this SG.
- (ii) The same polyester graft was used for all other SGs. The in-plane orthotropic elastic behaviour of this fabric was characterized in a previous study²². Bending rigidity of shell

elements used to mesh the PET fabric was adjusted according to a procedure detailed in this study. Parameters of this model are reported in table 6. This orthotropic model was implemented in the Abaqus software by using a “Lamina” material model.

Stent-graft modelling specificities

Compared to other SGs, the generation of Ta-SG, An-SG and Ex-SG models was particularly complex. This section presents the modelling details of these three SGs.

Talent (Ta-SG)

The metal structure of Ta-SG consists of five Z-stents as well as a longitudinal bar which increases longitudinal rigidity (or columnar strength) of the device in order to reduce the risk of SG migration. This bar has been modelled, meshed and added within the metal structure of the Ta-SG numerical model (see figure 3).

Anaconda (An-SG)

This SG has a characteristic “accordion” shape due to the way the stents are sewn onto the graft. Thus, modelling the crimped geometry of the fabric was particularly complex. The adopted approach to obtain this geometry can be divided in three steps (see figure 4):

- (i) Circular stents were prestressed: a sinusoidal longitudinal displacement was imposed to the stents centreline in order to give them a “crisp” shape (the amplitude and the number of periods of this sinusoidal displacement was measured on the stents of a stretched An-SG sample),
- (ii) Prestressed “crisp” stents were then fixed on the cylindrical textile with the above-mentioned “tie constraint”,

(iii) The prestresses were then released, so that the stents came back almost entirely to their initial circular shape. Thus, deformation equilibrium between stents and graft was reached, resulting in the “accordion” shape of the textile.

Furthermore, the following assumption was made concerning the stent wire. Each ring stent consists of a very thin Nitinol wire of 0.05 mm radius rolled several times in a concentric fashion. A single NiTi wire with equivalent radius of 0.15 mm was considered. This radius was calculated in order to obtain the same bending rigidity as the actual wire, while tensile properties were kept identical.

Excluder (Ex-SG)

For this device, the major difficulty laid in the particular fixation of the stent to the graft (as shown in figures 1 and 5): the stent is encapsulated between the graft and a thin polymeric strip. This type of fixation allows stent slight translation in the longitudinal direction. No equivalent algorithm was found in Abaqus to approximate this type of fixation. Consequently, the thin helical strip was modelled, preformed and applied against both stent and graft in order to encapsulate and allow the stent to translate longitudinally (see figure 5). A kinematic bonding (“tie constraint”) was applied between the graft outer surface and the strip inner surface in order to avoid motion between these two components. Neither bonding nor friction was considered between the stent and the textile and between the stent and the helical strip.

Simulation of SG bending and intraluminal pressurisation

Boundary conditions

To assess their mechanical performances, the above mentioned SG limbs were subjected to a severe bending followed by an intraluminal pressurisation (see figure 6). This type of boundary condition was chosen in order to mimick the *in vivo* deformations undergone by the SG in extremely tortuous AAA. The simulation consisted of three steps: (i) SG bending, (ii) adjustment of the distance between SG tips and (iii) intraluminal pressurisation.

(i) As previously reported^{22,23}, each SG tip was considered a rigid body controlled by a reference point (RP1 and RP2) (see figure 6). Opposite rotations were applied onto RP1 and RP2 about the x -axis, until an angle α of 90° and 180° was reached. The other two rotations were locked in order to maintain the SG in the yz -plane. Rigid body motions were prevented by locking the translations along the x and y axes. In order to avoid spurious tension in the longitudinal direction, the translation along the z -axis was left free.

(ii) Once SG bending was completed, distance between RP1 and RP2 was adjusted to reach 70 mm for $\alpha = 90^\circ$ and 35 mm for $\alpha = 180^\circ$, by applying opposite translations along the z -axis. These boundary conditions allowed SGs to be compared in identical loading conditions (figure 6).

(iii) Finally, each SG was subjected to a pressure of 150 mmHg applied on the inner surface of the graft. During the entire simulation, fabric porosity was not taken into account.

Numerical specificities

As this type of simulation involved complex geometric, material and, especially, contact nonlinearities, an explicit scheme was preferred. The ratio between global kinetic and strain energies was kept to a maximum of 5 - 10% in order to remain in a quasi-static case^{27,28}.

Assessment criteria of SG mechanical performances

In this study, mechanical performances of SGs included both flexibility and the mechanical response of each component to the loading conditions. Assessment of these performances was based on the following criteria which were previously described²².

Luminal reduction rate (LR)

Flexibility was evaluated by the calculation of *LR*. *LR* of SG cross-section was defined as the reduction of SG cross-sectional area between initial ($S_0 = \pi R_G^2$ corresponding to $\alpha = 0^\circ$, with R_G the initial graft radius, see table 2) and deformed state (S corresponding to $\alpha = 90^\circ$ or $\alpha = 180^\circ$):

$$LR = 100 \left(1 - \frac{S}{S_0}\right) \quad (\%) \quad (1)$$

This criterion characterized the variation of SG cross-sectional area. The maximal *LR* (LR_{max}) was defined as the highest value obtained among the 100 cross-sections observed for a given value of α . LR_{max} was then plotted for each SG at $\alpha = 90^\circ$ and $\alpha = 180^\circ$. A clinically relevant threshold value of LR_{max} was defined at 60 %, according to our surgical team experience.

Stresses in stent (σ_S^{max})

Maximal Von Mises stress (σ_S^{max}) in the deformed stents was derived from Abaqus numerical results. This criterion took into account tension/compression, bending as well as torsion of the stents. σ_S^{max} was also calculated for intraluminal pressure of 75 mmHg which corresponded to the diastolic pressure. Therefore it was possible to calculate $\Delta\sigma_S$ which corresponded to the variation of σ_S^{max} between 75 and 150 mmHg.

Strains in fabric (ϵ_{LG} and ϵ_{CG})

Longitudinal membrane strain (ϵ_{LG}) and circumferential membrane strain (ϵ_{CG}) were calculated by averaging out values of membrane strains for inner and outer surfaces of the shell elements. For that purpose, a local coordinate system (\vec{e}_L, \vec{e}_C) was defined along the yarn directions in order to ensure that output values corresponded to ϵ_{LG} and ϵ_{CG} .

The three above mentioned criteria were calculated for $\alpha = 90^\circ$ and $\alpha = 180^\circ$. Matlab R2011a was used to post-process the results obtained from Abaqus. Results were classified according to stent shape.

RESULTS

SG global deformation

Figure 7 represents SG deformed geometries for corresponding values of α (90° and 180°). No major SG kink was observed for $\alpha = 90^\circ$, except for the Ta-SG. For $\alpha = 180^\circ$, there were some significant differences between devices. Ze, Ta, and En-SGs displayed major kinks in their central area where the stents collapsed and dragged down the fabric with them. On the contrary, deformation of other devices was homogeneous along their entire length. Because of its “accordion” shape, An-SG unfolded and subsequently got longer when intraluminal pressure was applied.

Maximal luminal reduction rate (LR_{max})

Figure 8A shows LR_{max} plotted for each device at $\alpha = 90^\circ$ and $\alpha = 180^\circ$.

At $\alpha = 90^\circ$, all SGs except Ta-SG had LR_{max} equal or less than 20%.

At $\alpha = 180^\circ$, two groups of SGs could be identified. The first group included Z-stented SGs (Ta, En, Ze and Zlp-SGs). LR_{max} of these SGs was high, reaching peak values between 70 and 80% (Ta, En, Ze-SGs). For Zlp-SG, LR_{max} peak value remained lower than 40%. The second group included the other devices which had either spiral stents (Ao, Ex and Zs-SGs) or separated circular stents (An-SG). LR_{max} values in this group were low (less than 25%).

Because these values suggested possible phase transformations for some SGs, a finer analysis in terms of principal stresses was also performed. This analysis revealed that minimal principal stresses -negative values, hence in compression- could drop below -400 MPa for Ta and En, with respective values close to -500 and -600 MPa. However, the maximal principal stresses -hence in traction- never exceeded 390 MPa for any SG.

Stresses in the stents (σ_S^{max})

Calculated values of σ_S^{max} are presented in figure 8B for each SG at $\alpha = 90^\circ$ and $\alpha = 180^\circ$.

At $\alpha = 90^\circ$, σ_S^{max} was less than 300 MPa for all SGs except Ta and Ex-SGs. The highest σ_S^{max} were recorded for Ta-SG, with a peak value of 560 MPa.

At $\alpha = 180^\circ$, two groups of SGs could be identified. The first group included Z-stented SGs (Ta, En, Ze and Zlp-SGs) except Ex-SG which had a Z-spiral stent. In this group, σ_S^{max} values were greater than 300 MPa. The second group included spiral and circular-stented SGs (Ao, An and Zs-SGs). In this group, σ_S^{max} values were lower than 300 MPa.

For all SGs, highest σ_S^{max} were located at the level of stent apex.

$\Delta\sigma_S$ at 90° and 180° ranged from 0 (Ta-SG) to 31 MPa (Ao-SG). This corresponds to very small strain magnitude $\Delta\varepsilon_S$ ranged from 0 to 0.08%.

Strains in the fabric (ε_{LG}^{max} and ε_{CG}^{max})

Figure 8C (8D, respectively) represents ε_{LG}^{max} (ε_{CG}^{max} , respectively) plotted for each device at $\alpha = 90^\circ$ and $\alpha = 180^\circ$.

For SGs with PET fabric, ε_{LG}^{max} was higher than ε_{CG}^{max} for the same angulation. For all SGs except Ex-SG, ε_{LG}^{max} and ε_{CG}^{max} were low ($\varepsilon_{LG}^{max} < 7\%$ and $\varepsilon_{CG}^{max} < 2\%$) at 90° and 180° . For Ex-SG, ε_{LG}^{max} and ε_{CG}^{max} were up to 18% for $\alpha = 180^\circ$ (the fabric of this SG was ePTFE).

Maximum strains were mainly located at the inner curvature of the SG, particularly between stents or between stent patterns for SGs with a single stent. In these areas, stents dragged the fabric down with them and sometimes overlapped causing important local fabric stretches. For Ex-SG, the maximum strain in the fabric was located at the interface between the graft and the helical strip.

DISCUSSION

The present study was a comparison of the mechanical performances of eight currently marketed SGs. The results confirmed that the stent geometry strongly influences SG flexibility and mechanical stresses in stents.

Flexibility (LR_{max})

At 90°, no significant difference was observed in the LR_{max} value between the different devices except for Ta-SG. LR_{max} value (20%) remained well below the clinical threshold of 60% commonly associated with limb thrombosis or symptomatic kinks. Therefore, these results confirm that most current SGs could be used safely in iliac angulations up to 90°.

At 180°, circular, spiral or Z-spiral stents provided better flexibility than Z-stents. Interestingly, optimization of Z-stents such as with Zlp-SG (decreased number of Z-periods and shorter stents) was associated with lower LR_{max} values. For this particular SG, better stent interlocking was observed during bending which resulted in a lower luminal reduction.

Stresses in stents / Fatigue

At 90°, stresses in stents were higher for Ta-SG. It was possible to identify from the model that the highest stress was localized in the longitudinal bar even for low angulations. Furthermore, it was possible to demonstrate that pressurisation was responsible for the particular V-shape observed with this SG at 90° (figure 7). The high stress undergone by Ex-SG ($\sigma_s^{max} = 375$ MPa) at 90° may be explained by the observed local wrinkling of both graft and strip which caused the stent to jam at the level of the angulation.

At 180°, the highest stresses were observed with Z-stents. Lower stresses of spiral and circular stents were associated with minimal stent deformation even for the highest angulation.

NiTi stents remained in their linear elastic domain during simulation, since σ_S^{max} never reached the stress required to induce the forward martensitic transformation ($\sigma_L^S = 390$ MPa). For Ta-SG and En-SG, higher σ_S^{max} values are obtained due to compression stresses as was shown with the principal stress analysis. Traction stresses remain below the forward martensitic transformation. Figure 8B suggests that stresses in 316L stainless steel Z-stents of the Ze-SG ($\sigma_S^{max} = 740$ MPa) remain well below the yield stress of this alloy ($\sigma_e = 1550$ MPa) and therefore plasticity was never reached during simulation.

Stress ($\Delta\sigma_S$) and ($\Delta\varepsilon_S$) strain variations within the stents between diastolic (75 mmHg) and systolic (150 mmHg) pressures were small for all SGs. Accounting for the calculated maximal Von Mises stresses σ_S^{max} , such very small mechanical oscillations around σ_S^{max} should not be detrimental to the fatigue life of the stents²⁹. These data are consistent with the fact that stent fracture occur rarely in clinical practice with the SGs considered in this study.

Strains in fabric

At 90° and 180°, strains in fabric were well below ultimate strains of graft materials (around 20% for PET and 40% for ePTFE, see tables 5 and 6). This is consistent with the fact that fabric tear rarely occur with the devices considered in this study.

FEA models allowed to assess areas of maximal strain in the fabric. Particularly, Z-stents were associated with the highest strains in the fabric compared to circular and spiral stents.

The ePTFE fabric of Ex-SG was much more deformed than the PET textile of other SGs. Strains undergone by this material were higher since its Young's modulus was much lower (55.2 MPa) than those of the PET fabric ($E_L = 225$ MPa and $E_C = 1000$ MPa).

Limitations

Several limitations of this study may be mentioned.

Friction between stents and fabric caused by micro motion of stents despite sutures was not considered. This phenomenon may cause localized fabric wear and tear as previously reported⁵.

Sutures were not modelled but approximated by a bonding algorithm between stents and graft in order not to increase computational time dramatically.

Isotropic linear elastic constitutive law of ePTFE fabric was taken from the single numerical study which used this material²¹. However, isotropy is maybe not representative of the actual mechanical behaviour of this material. Preferential orientations of ePTFE microstructure were observed in the study of Catanese et al. (1999)²⁶. Because no fabric sample was available to us, it was impossible to perform proper characterization of ePTFE mechanical behaviour.

Blood flow and corresponding shear were not considered as FSI simulations would have been much more complex to implement. For the same reason, interaction between SG and arteries were not computed.

Perspectives

This report is the first step of a global study on the mechanical behaviour of aortic SGs. Further computations are underway and aim to model bifurcated SGs. SG deployment in aortic numerical models is also being developed. The next step would consist in simulating

SG deployment within patient-specific AAAs²⁴. Moreover, a future study could focus on optimizing the suture system by using our simulations in a multi-scale analysis.

Another application of this technology could be the optimization of SG design. Mechanical performances of newly designed SGs could be tested numerically without the need for prototypes and bench-tests.

CONCLUSION

This study confirmed that stent design strongly influences mechanical performances of aortic stent-grafts. Spiral and circular stents provide greater flexibility and lower stress values than Z-stents.

REFERENCES

1. Albertini JN, Macierewicz JA, Yusuf SW, et al. Pathophysiology of proximal perigraft endoleak following endovascular repair of abdominal aortic aneurysms: a study using a flow model. *Eur J Vasc Endovasc Surg.* 2001;22:53-56.
2. Baum RA, Stavropoulos SW, Fairman RM, et al. Endoleaks after endovascular repair of abdominal aortic aneurysms. *J Vasc Interv Radiol.* 2003;14:1111–1117.
3. Carroccio A, Faries PL, Morrissey NJ, et al. Predicting iliac limb occlusions after bifurcated aortic stent grafting: Anatomic and device-related causes. *J Vasc Surg.* 2002;36:679–684.
4. Cochenec F, Becquemin JP, Desgranges P, et al. Limb graft occlusion following EVAR: Clinical pattern, outcomes and predictive factors of occurrence. *Eur J Vasc Endovasc Surg.* 2007;34:59–65.
5. Chakfe N, Dieval F, Riepe G, et al. Influence of the textile structure on the degradation of explanted aortic endoprostheses. *Eur J Vasc Endovasc Surg.* 2004;27:33–41.
6. Heintz C, Riepe G, Birken L, et al. Corroded nitinol wires in explanted aortic endografts: An important mechanism of failure? *J Endovasc Ther.* 2001;8:248–253.
7. Zarins CK, Arko FR, Crabtree T, et al. Explant analysis of AneuRx stent grafts: Relationship between structural findings and clinical outcome. *J Vasc Surg.* 2004;40:1–11.
8. Bartoli MA, Thevenin B, Sarlon G, et al. Secondary Procedures After Infrarenal Abdominal Aortic Aneurysms Endovascular Repair With Second-Generation Endografts. *Ann Vasc Surg.* 2012;26:166-174.
9. Albertini JN, DeMasi MA, Macierewicz J, et al. Aorfix stent graft for abdominal aortic aneurysms reduces the risk of proximal type 1 endoleak in angulated necks: Bench-test study. *Vasc.* 2005;13:321-326.
10. Weale AR, Balasubramaniam K, Hardman J, et al. Use of the Aorfix (TM) stent graft in patients with tortuous iliac anatomy. *J Cardiovasc Surg.* 2010;51:461-466.
11. Morris L, Delassus P, Walsh M, et al. A mathematical model to predict the in vivo pulsatile drag forces acting on bifurcated stent grafts used in endovascular treatment of abdominal aortic aneurysms (AAA). *J Biomech.* 2004;37:1087-1095.
12. Howell BA, Kim T, Cheer A, et al. Computational fluid dynamics within bifurcated abdominal aortic stent-Grafts. *J Endovasc Ther.* 2007;14:138-143.

13. Figueroa CA, Taylor CA, Yeh V, et al. Effect of Curvature on Displacement Forces Acting on Aortic Endografts: A 3-Dimensional Computational Analysis. *J Endovasc Ther.* 2009;16:284–294.
14. Figueroa CA, Taylor CA, Yeh V, et al. Preliminary 3D computational analysis of the relationship between aortic displacement force and direction of endograft movement. *J Vasc Surg.* 2010;51:1488–1497.
15. Li Z, Kleinstreuer C. Analysis of biomechanical factors affecting stent-graft migration in an abdominal aortic aneurysm model. *J Biomech.* 2006;39:2264–2273
16. Li Z, Kleinstreuer C. Computational analysis of type ii endoleaks in a stented abdominal aortic aneurysm model. *J Biomech.* 2006;39:2573–2582.
17. Li Z, Kleinstreuer C, Farber M. Computational analysis of biomechanical contributors to possible endovascular graft failure. *Biomech Model Mechanobiol.* 2005;4:221–234.
18. Li ZH, Kleinstreuer C. Blood flow and structure interactions in a stented abdominal aortic aneurysm model. *Med Eng Phys.* 2005;27:369–382.
19. Molony D, Callanan A, Kavanagh E, et al. Fluid-structure interaction of a patient-specific abdominal aortic aneurysm treated with an endovascular stent-graft. *Biomed Eng Online.* 2009;8:24–24.
20. Molony DS, Kavanagh EG, Madhavan P, et al. A computational study of the magnitude and direction of migration forces in patient-specific abdominal aortic aneurysm stent-grafts. *Eur J Vasc Endovasc Surg.* 2010;40:332–9.
21. Kleinstreuer C, Li Z, Basciano CA, et al. Computational mechanics of Nitinol stent grafts. *J Biomech.* 2008;41:2370–2378.
22. Demanget N, Avril A, Badel P, et al. Computational comparison of the bending behavior of aortic stent-grafts. *J Mech Behav Biomed Mater.* 2012;5:272-282.
23. Demanget N, Latil P, Orgeas L, et al. Severe bending of two aortic stent-grafts: an experimental and numerical mechanical analysis. *Ann Biomed Eng.* 2012. DOI: 10.1007/s10439-012-0618-0.
24. De Bock S, Iannaccone G, De Santis G, et al. Virtual evaluation of stent graft deployment: a validated modeling and simulation study. *J Mech Behav Biomed Mater.* 2012;13:129-139.
25. Auricchio F, Taylor RL. Shape-memory alloys: Modelling and numerical simulations of the finite-strain superelastic behavior. *Comput Meth Biomech Biomed Eng.* 1997;143:175-194.
26. Catanese J, Cooke D, Maas C, et al. Mechanical properties of medical grade expanded polytetrafluoroethylene: The effects of internodal distance, density, and displacement rate. *J Biomed Mater Res.* 1999;48:187-192.

27. Kim J, Kang YH, Choi HH, et al. Comparison of implicit and explicit finite-element methods for the hydroforming process of an automobile lower arm. *Int J Adv Manuf Tech.* 2002;20:407–413.
28. Auricchio F, Conti M, De Beule M, et al. Carotid artery stenting simulation: From patient-specific images to finite element analysis. *Med Eng Phys.* 2011;33:281-289.
29. Robertson SW, Pelton AR, Ritchie RO. Mechanical fatigue and fracture of Nitinol. *Int Mater Rev.* 2012;57:1-36.

TABLES

Table 1 **Manufacturing features of considered SG limbs**

	Stent material	Stent shape	Graft material	Attachment of stents to graft
<i>Aorfix</i>	Nitinol	Spiral	Polyester	Sutures
<i>Anaconda</i>	Nitinol	Circular	Polyester	Sutures
<i>Endurant</i>	Nitinol	Z	Polyester	Sutures
<i>Excluder</i>	Nitinol	Combination spiral/Z	ePTFE	Stent encapsulation ^(*)
<i>Talent</i>	Nitinol	Z + additional side bar	Polyester	Sutures
<i>Zenith Flex</i>	316L Stainless Steel	Z	Polyester	Sutures
<i>Zenith Spiral-Z</i>	Nitinol	Combination spiral/Z	Polyester	Sutures
<i>Zenith LP</i>	Nitinol	Z	Polyester	Sutures

^(*) Stent encapsulated between the graft and a thin polymeric strip

Table 2 Geometrical and computational features of SG limbs

	Ao-SG	An-SG	En-SG	Ex-SG	Ta-SG	Ze-SG	Zs-SG	Zlp-SG
<i>Total number of elements</i>	71928	74033	89892	242192	71560	65460	56912	63820
Graft								
Radius R_G (mm)	7.867	7.842	7.792	7.862	7.742	7.852	7.792	7.822
Length L_G (mm)	88.0	92.9	91.5	94.9	108.1	99.4	90.7	92.9
Stents								
Stent height H_S (mm)	L_G	0.40	8.3	5.5/13.5	14.6	11.7	7.5	8.6
Number of stents N_S	1	18	9	1	6 + side bar	6	1	8
Wire radius R_S (mm)	0.125	0.15	0.20	0.13	0.25	0.14	0.20	0.17

Table 3 **Material properties of 316L stainless steel**

Parameters	Values
Young's modulus E (MPa)	210000
Poisson's ratio ν	0.3
Yield stress σ_e (MPa)	1550
Ultimate tensile strength σ_R (MPa)	> 2300

Table 4 **Material properties of Nitinol** (Kleinstreuer et al., 2008)

Parameters	Values
Austenite elasticity E_A (MPa)	40000
Austenite Poisson's ratio ν_A	0.46
Martensite elasticity E_M (MPa)	18554
Martensite Poisson's ratio ν_M	0.46
Transformation strain ε^L	0.04
Start of transformation loading σ_L^S (MPa)	390
End of transformation loading σ_L^E (MPa)	425
Ultimate tensile strength σ_R (MPa)	827 – 1172

Table 5
2008)

Material properties of ePTFE (Catanese et al., 1999; Kleinstreuer et al.,

Parameters	Values
E (MPa)	55.2
ν	0.46
Ultimate strain ε_R	0.2 - 0.477

Table 6 **Material properties of PET**

Parameters	Values
$E_{\theta=0^\circ} = E_L$ (MPa)	$225 \pm 10\%$
$E_{\theta=90^\circ} = E_C$ (MPa)	$1000 \pm 10\%$
ν_{LC}	0.2
G (MPa)	3.6
ε_R^L	0.23
ε_R^C	0.18
D_L (10^{-4} N.mm)	4.0
D_C (10^{-4} N.mm)	18

E_L : longitudinal Young's modulus. E_C : circumferential Young's modulus. ν_{LC} : Poisson's ratio. G : shear modulus. ε_R^L : longitudinal ultimate strain. ε_R^C : circumferential ultimate strain. D_L : longitudinal bending stiffness. D_C : circumferential bending stiffness.

LEGENDS

Figure 1 - Marketed stent-graft limbs and corresponding numerical models.

Figure 2 - Generation of stent mesh.

Figure 3 - Numerical modelling of the metal structure of the Ta-SG.

Figure 4 - Outline of the different steps necessary for the implementation of the An-SG numerical model.

Figure 5 - Numerical model of the Ex-SG: Encapsulation of the stent between the graft and the thin helical strip.

Figure 6 - Schematic view of SG in corresponding boundary conditions: angulation of 90° (left hand side) and 180° (right hand side)

Figure 7 - Deformed SGs for $\alpha = 90^\circ$ and $\alpha = 180^\circ$

Figure 8 - Quantitative assessment of SGs mechanical performances for $\alpha = 90^\circ$ and $\alpha = 180^\circ$: (A) LR_{max} , (B) σ_S^{max} , (C) ε_{LG}^{max} , (D) ε_{CG}^{max} .