VIRTUAL STENT GRAFT DEPLOYMENT IN PATIENT-SPECIFIC ABDOMINAL AORTIC ANEURYSMS: FINITE ELEMENT RESULTS

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SUMMARY
The presented work details the computer simulation of the deployment of a stent graft in a patient-specific Abdominal Aortic Aneurysm (AAA) model, using the finite element method. The entire procedure is modeled, with the stent graft being crimped and bent according to the vessel geometry, and released in two steps. Deformations, external and internal forces during the procedure can be quantified and visualized and related to relevant stent graft characteristics. Follow up studies will include experimental validation and model extension.

INTRODUCTION
An AAA is a local dilatation of the aorta, commonly involving the region between the renal arteries and the aortic bifurcation. Rupture of this weakened aortic tissue results in life threatening internal bleeding, with a very high mortality rate. Conventional treatment is open surgical repair (OSR), a high risk invasive procedure. An alternative to OSR is endovascular repair with stent-graft implantation, which has proven to be a valid alternative, with clear benefits in terms of length of hospital stay, reduced operational mortality and less trauma [1]. In the long term however, the outcome does not differ significantly due to stent graft related complications, including graft migration, endoleakage and fatigue failings.

Numerical modeling of stent grafts and aneurysms can be used to assess the strains, stresses and forces occurring in vivo and play an important role in research on stent graft and vessel mechanics [2]. In a recent study, Kleinsteuer et al. [3] investigated a tubular, diamond-shaped stent graft under cyclic loading and its deployment in a circular AAA neck. To the best of our knowledge, no prior work has been published on virtual stent graft deployment in a patient-specific AAA model. In this study, the deployment of a stent design resembling the Medtronic Talent stent graft (Figure 1) in a patient-specific AAA model is simulated.

METHODS
The nickel titanium alloy’s mechanical properties were obtained from literature [3]. The super elastic behavior of the nitinol material is modeled using an available user-defined material subroutine, based on the model as described by Auricchio and Taylor [4]. Stent geometry and dimensions were reconstructed from the Talent device manual [5] and direct measurements on a sample with a caliper. From these dimensions, a structured hexahedral mesh was constructed using the open-source software pyFormex [6], by sweeping a centerline with a cross-section of twelve quadrilateral elements, approximating the circular wire. This method results in a structured mesh, and allows for easy nonuniform refining. The unloaded device has an outer diameter of 24 mm (proximal main body) and 14 mm (distal iliac leg), and is shrunk to a minimal diameter of 6.8 mm upon insertion. The AAA geometry has a proximal aorta neck and distal iliac artery diameter of an estimated 16 mm and 11 mm, which means that in this particular case the stent graft is larger than advised by the oversizing guidelines [5]. The short iliac leg (see Figure 1) is not taken into account.

Figure 1: Medtronic Talent abdominal stent graft [6]

The AAA model geometry was previously used in FSI research investigating the role of intraluminal thrombus on the rupture risk in AAA patients [7]. As in [7], the vessel wall was assumed to have a uniform thickness of 2 mm, though it was re-meshed with first order hexahedral (brick) elements using an in-house developed algorithm within pyFormex [8] (thrombus was excluded). It has been shown that hexahedral elements achieve better mesh convergence and accuracy [9]. The final mesh consisted of 36288 elements.

The aortic wall material was assumed to be hyperelastic, incompressible, isotropic, and homogenous, and was based on the constitutive model proposed by Raghavan and Vorp [10]. The model parameters were set to $C_{10} = 17.4$ N/cm² and $C_{20} = 188.1$ N/cm², being the mean values obtained from uniaxial testing of excised AAA specimens [10].

As the device is not expanded beyond its original diameter during the deployment, it is assumed that the polyester graft material has no effect on the deformation of the stent, and it is not incorporated into the model. For the actual sample, the nitinol wire of the main body and iliac leg are sewn to this continuous polyester graft. In the computer model, this connection is made using behavioral connector elements, preventing the extension of the segments, while allowing the springs to move towards each other (free compression of the fabric).
A deformable cylinder is used to perform the crimping and bending of the stent graft through contact forces, mimicking crimping and stent insertion. The finite element calculations are performed with Abaqus/Standard (SIMULIA, Dassault Systèmes).

Figure 2: AAA stent graft deployment: a) expanded, strain-free situation; b) crimping and c) bending of the stent graft; d) release of main body; e) fully deployed device.

RESULTS AND DISCUSSION

The present study shows the implantation of a stent graft in a patient-specific AAA model. In a first step, the main body and iliac leg of the stent are crimped to the same diameter, simulating the crimping into a catheter (figure 2.b). The self-expanding stent is bent according to the geometry of the receiving vessel (2.c). The deformable cylinder (not depicted) releases the stent graft in two steps, first allowing the main body to make contact with the proximal aneurysm neck (2.d) after which the iliac leg expands into the iliac branch (2.e). Contact between the stent and the AAA model is only activated in these last steps.

The evolution of stresses and strains can be assessed from the simulation. Figure 3.a shows the Von Mises stress that remains in the stent after deployment, with a maximum of 569.1 MPa. At this time the stent graft exerts an outward force, one of the mechanisms which keeps the device in place. This force totals 4.72 N in the proximal neck, and 4.37 N in the iliac artery. When comparing configurations, these figures can help gauge the resistance to stent migration, or the risk for type I endoleak (leakage from the attachment sites). Figure 3.b shows the effect of the contact force on the aorta. In this simulation, the distention remains limited, with maximal logarithmic strain of 3.09%.

CONCLUSIONS

Computer modeling increasingly plays an important role in the design process of medical devices. This study shows a method of calculating internal and external forces during stent graft insertion, and all relevant derived parameters. In future studies, this will allow to compare the performance and behavior of different stent graft configurations in various patient-specific AAA geometries. Follow up studies will recreate stent graft geometry more accurately by using µCT scans of existing devices.

Figure 3: a) Von Mises stress in the stent and b) logarithmic strain in the AAA wall at the end of the deployment procedure.

The numerical results will be validated by performing radial stiffness tests on the stent grafts and in experimental set-ups using silicone AAA models. The model will be extended to also include the polyester graft, which will allow to employ a cyclic arterial pressure to the internal surface of this graft, enabling the numerical evaluation of the fatigue life of stent grafts in a patient-specific geometry.

REFERENCES

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