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A computational study of stent performance by considering vessel anisotropy and residual stresses

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ABSTRACT

Finite element simulations of stent deployment were carried out by considering the intrinsic anisotropic behaviour, described by a Holzapfel–Gasser–Ogden (HGO) hyperelastic anisotropic model, of individual artery layers. The model parameters were calibrated against the experimental stress–stretch responses in both circumferential and longitudinal directions. The results showed that stent expansion, system recoiling and stresses in the artery layers were greatly affected by vessel anisotropy. Following deployment, deformation of the stent was also modelled by applying relevant biomechanical forces, i.e. in-plane bending and radial compression, to the stent–artery system, for which the residual stresses generated during deployment were particularly accounted for. Residual stresses were found to have a significant influence on the deformation of the system, resulting in a re-distribution of stresses and a change of the system flexibility. The results were also utilised to interpret the mechanical performance of stent after deployment.

1. Introduction

Stents have been extensively used to treat obstructions of human blood vessels, such as the narrowing of coronary arteries due to plaque accumulation and the formation of aneurysms and blockage caused by anomalous vessel deformation [4]. Ideal process of stent deployment requires an expansion which is sufficient to clear vessel obstructions and also induces minimal damage to the vessel walls [3], as in-stent restenosis can be triggered by unsuccessful treatments including vessel overstretch and injury, incomplete expansion of the stent and fracture of the stent struts. Karimi et al. [12] assessed arterial vulnerability during stenting and confirmed a risk of damage to the arterial walls, especially when covered with calcified plaque. Computational modelling has been widely used to study the interactions between stents and blood vessels during and after stent deployment, especially finite element (FE) analyses which can model the full process of stent expansion effectively and provide the details of stress distribution within the stent–artery system.

The state of the art in the characterisation of arterial wall layers suggests that the fibre-reinforced structure of intima, media and adventitia layers, forming the vessel wall, must be taken into account when modelling the deformation of blood vessels. Holzapfel et al. [7] first showed how the collagenous components of the vessel layers can be modelled by means of fibre orientation to introduce the anisotropic material properties. Their results were in good agreement with the available data for arteries under axial extension, radial inflation and torsion. The same framework has been later employed to model the layer-specific mechanical properties of human coronary arteries [8]. The study also analysed 13 human nonstenotic left anterior descending (LAD) coronary arteries. The arterial walls were separated into three layers whose mechanical properties were tested in both longitudinal and circumferential directions. In all cases, large difference was found between the patients and also the three layers exhibited highly non-linear anisotropic behaviour between the longitudinal and circumferential directions. The computational model showed good agreement with the experimental results, confirming its capability of describing the mechanical behaviour of such materials. However, the implementation of arterial anisotropy in finite element modelling of stent biomechanics is still very limited. Existing work only adopted simplified model to model stent expansion, such as the single-layer anisotropic model used in Auricchio et al. [2] and the redundant HGO anisotropic model (without the consideration of fibre family) used in Schiavone and Zhao [19]. Consequently, the results were less conclusive and further studies are required to quantitatively elucidate the effects of vessel anisotropy on stent deployment.

Body movements, tissue tethering and blood flow are key factors that induce mechanical loading on arterial branches. When the mechanical stability of the vessel is lost, buckling can occur which generates large bending deformations on the artery [6]. As reviewed by Fortier et al. [4], implanted stents may undergo a variety of repetitive deformations caused by muscular movements during daily activities such as walking, flexing, respiration and heartbeat. Smouse et al. [17] studied the stented femoropopliteal arterial segment in 14 limbs of 7 human cadavers. The limbs were flexed to mimic movements such as walking, climbing stairs, sitting and standing. Bending, associated with compression, was observed for arteries behind the knee, and the presence of stents increased the axial rigidity of the vessel, which can cause stent

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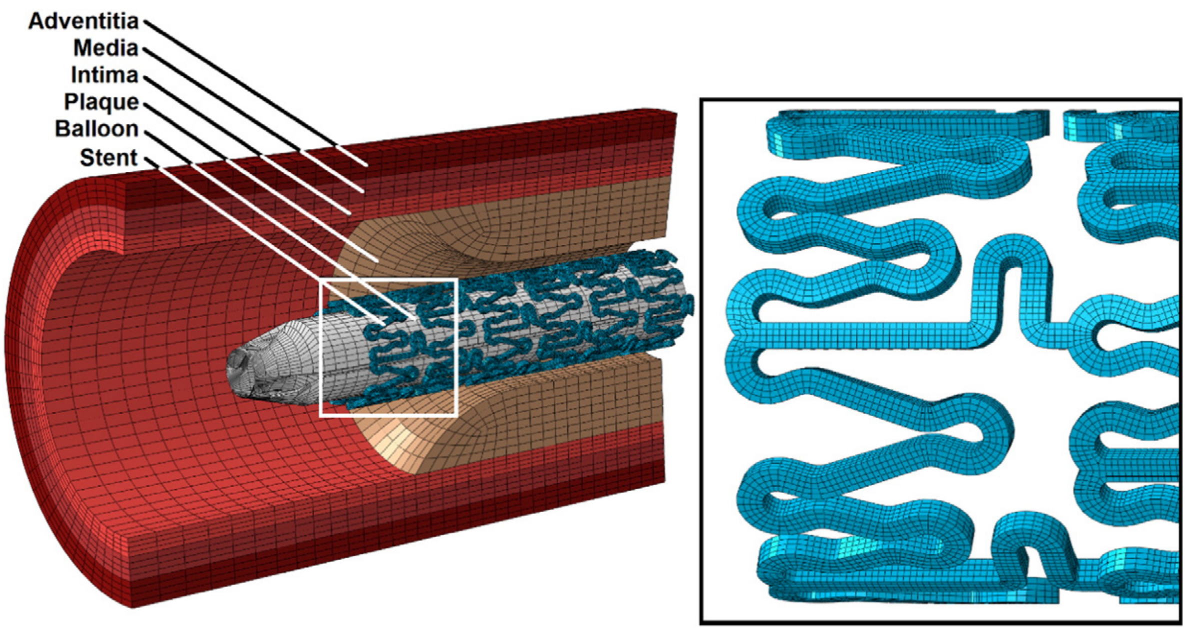

Fig. 1. Finite element mesh for the stent–artery–balloon system.

kinking or fracture. Rosenfield et al. [15] studied stented arterial branches in various vascular sites that were subjected to radial compression, such as six superficial femoral arteries and five haemodialysis conduits. Results showed that restenosis occur as a result of stent compression, and two mechanisms of compression were identified, namely, two point compressive force and complete circumferential encroachment. Despite redilation of the stents, restenosis occurred from recurrent compression in most sites. Also, radial compression of the artery can be caused by external pressures due to cardiac contraction or stenosis. Consequently, it is of importance to evaluate stent performance under biomechanical forces, especially bending and radial compression, which has not been well studied yet. In addition, balloon-expandable stents rely on large plastic deformation to resist recoiling and keep the blocked artery open, which inevitably introduces severe residual stresses in the stents. The residual stresses tend to locate at the severely deformed U-bend regions, with magnitude close to the ultimate tensile strength of the materials [16]. Effect of these residual stresses on stent performance is also considerably lacking in literature.

In this paper, expansion of Xience stent inside an artery with stenosis was simulated using finite element analyses. The work compared the behaviour of stent expansion in an artery described by isotropic and anisotropic hyperelastic models, respectively. After deployment, two major biomechanical movements of the stent–artery system were considered, i.e., bending and radial compression. In particular, the effects of residual stresses on stent deformation were analysed by comparing simulations with and without accounting for the residual stress states. This work highlighted the importance of vessel anisotropy and residual stresses in assessing stent performance during and after deployment.

2. Methodology

2.1. Geometrical models and meshes

Geometrical model for Xience stent was generated using Abaqus CAE [1]. The stent was 12 mm long, with a strut thickness of 80 μm and a crimped diameter of 1.58 mm. A tri-folded balloon with a total length of 16 mm was generated using NX (Siemens PLM Software, UK). The major section of the folded balloon was 14 mm long, with a diameter of 1.24 mm. The ends of the balloon were modelled as a simple circle with a diameter of 0.7 mm (matching the diameter of the angioplasty catheter), and gradually changed to the fully folded section over a length of 1 mm. The artery was 40 mm long, with a diameter of 4 mm and a wall thickness of 1 mm. The wall was divided into three layers, i.e., intima, media and adventitia, with respective thicknesses of 0.27 mm, 0.35 mm and 0.38 mm. The plaque was 10 mm long and its thickness was gradually increasing from zero to 1 mm over a distance of 2 mm at each side, corresponding to a maximum occlusion of 50%.

The stent was meshed using 8-node hexahedral elements with full integration and incompatible modes (C3D8I). This element formulation is suitable for modelling of large bending deformation which occurs during the stent expansion. All struts have four-layer elements through both the thickness and the width, resulting in 227,120 elements for the whole stent. The balloon was meshed using 4-node shell elements with reduced integration (S4R), and the total number of elements was 12,054. The artery–plaque system was meshed using 8-node hexahedral elements with reduced integration (C3D8R). There are four-layer elements through the thickness of each arterial layer and eight-layer elements through the thickness of the plaque. The total number of elements was 67,392 elements for the artery and 20,736 for the plaque. Mesh sensitivity study confirmed the numerical convergence for the mesh used in this study in terms of diameter change, recoiling and stress distribution for the system. Fig. 1 shows the geometry and the mesh for the whole balloon-stent–artery system.

2.2. Material models

Elastic–plastic behaviour was assigned to the Xience stent made of the Co–Cr alloy L605. The density, Young's modulus, yield stress and Poisson's ratio for the material are 9.1×10^{-6} kg/mm^3, 243 GPa, 476 MPa and 0.3, respectively. The hardening behaviour was modelled by expressing the increase of yield stress as a function of the plastic strain, obtained from the tensile stress–strain curve. A linear elastic model was adopted for the folded balloon, whose density, Young's modulus and Poisson's ratio are 1.1×10^{-6} kg/mm^3, 900 MPa and 0.3, respectively. The hypocellular plaque was modelled as an isotropic hyperelastic material using the Ogden strain energy potential, with parameters given in Zahedmanesh and Lally [18].

The Holzapfel–Gasser–Ogden (HGO) formulation was used to model the anisotropic behaviour of the arterial layers, with strain energy