Modeling of Stent Implantation in a Human Stenotic Artery

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Abstract—**The aim of this work is to introduce a methodology to study the stent expansion and the subsequent deformation of the arterial wall towards the outside direction in order arterial lesion to be rehabilitated and blood flow to be restored. More specifically, a coronary artery and the plaque are reconstructed using intravascular ultrasound and biplane angiography. The finite element method is used for the modeling of the interaction between the stent, balloon, arterial wall and plaque. Appropriate material properties and boundary conditions are applied in order to represent the realistic behavior of each component. We observe that stresses are increased at the region of the first contact between the stent and the wall, which may be considered crucial for plaque rupture. Furthermore, the average calculated stress on the plaque is higher than the average stress on the arterial wall. Thus, stent positioning and deployment depends on a considerable degree on the plaque properties rather than the general arterial geometry. Results indicate that numerical modeling can provide a prediction of the arterial behavior during stent implantation.**

Keywords—**Stent, Human artery, Atherosclerosis, Finite element method.**

I. INTRODUCTION

The cardiovascular disease affects the quality of life of patients and can lead to severe health problems or even death. Atherosclerosis involves arterial wall thickening and blood flow reduction, caused by the buildup of the plaque. Plaque consists of calcium, cholesterol crystals and other types of cells like smooth muscle cells. Plaque grows and causes artery occlusion [1]. Several different treatment techniques and procedures are available for rehabilitation including angioplasty, stent implantation and artery bypass. Angioplasty is an invasive procedure where a balloon catheter is placed into the artery, expands and re-opens the artery [2]. Angioplasty is not the best choice for all plaque types since lesions vary in composition, size and affected area. These difficulties, complications and short comings, which may presented during angioplasty, were faced using an invasive endovascular device, which is called stent.

A stent is a small metallic tube consisted of wires, being initially in a crimped condition, mounted on a balloon catheter device. Once in place, the balloon is inflated, the stent expands and compresses the arterial plaque reserving the inner artery wall open, after balloon removal. Like every mechanical procedure, there is a dependency on the mechanical properties, the geometry and the morphology of the involved components (artery, plaque, stent and balloon) [3].

Stent geometry and design could result in a different arterial behavior. Plaque types vary and are classified according to their stiffness. Thus, different plaque types demonstrate different response when they undergo the same stenting process.

The finite element method (FEM) can be used to study the interaction between the stent, the artery and the plaque components during stent placement. Stent and artery interaction was examined by Berry *et al*. [4]. The plaque component was also considered in more recent studies utilizing idealized arterial geometries such as stenosed straight [5-7] or curved [8] geometries. Arterial wall stresses caused by the stent implantation in stenotic arteries depend on the different plaque types [9]. A finite element model of a real patient artery was developed with the presence of stent, ignoring however the presence of plaque morphology [10].

In this study, the stent implantation procedure is simulated in a reconstructed patient's stenotic artery including the plaque component in the area of stenosis. The effect of the stent implantation device is represented in relation to stress distribution and deformation caused in the arterial wall and plaque.

II. MATERIAL AND METHODS

A. 3D Reconstruction

Data from a 62-year old smoker overweight male (BMI 27.8) with high levels of cholesterol in the blood

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(hypercho-lesterolemia) and hypertension problems were used. Intra-vascular ultrasound (IVUS) and angiography were employed. The reconstruction of the coronary artery was conducted adopting the approach introduced by Bourantas *et al.* [11]. This method establishes a particular segmentation methodology which detects the lumen and media adventitia borders and afterwards the detected borders are placed on the 3D catheter path, which is extracted by the processing of bi-plane angiography. The output of this procedure is two point clouds, which represent the arterial wall and lumen geometry. The plaque component is reconstructed after automated plaque characterization using the methodology presented in [12].

B. Computational Simulation

The 3D finite element model was developed in its unexpanded state and it consists of the artery, the plaque, the stent and the balloon. ANSYS 12.1 (ANSYS, Canonsburg, PA) was used for pre- and post-processing. Based on the Open Stent design [13], the geometry of the stent device was appropriately shaped for this specific human artery. The balloon was modeled as a plane cylinder and it was positioned in such a way that the inner surface of the stent was in initial contact with the outer surface of the balloon. The balloon-stent device was not in initial contact with the arterial lumen (Fig. 1). The plaque was lying in the inside area of the arterial wall (Fig. 1). The thickness of the plaque varied from 1.05mm to 1.40mm. The arterial segment was 17.54mm in length and the length of the plaque was 3.27mm. Figure 1 presents the whole model consisted of the artery, the plaque, the stent and the balloon in their initial condition. The mesh density was chosen based upon the elimination of the existing penetration between the contact pairs appearing in this model.

As far as modeling techniques are concerned, there are mainly two different approaches regarding the stent simulation procedure. The first method uses a consistent uniform pressure which increases and is placed directly in the inside surface of the stent or the balloon-stent device [14-16]. In the second method radial displacement is enforced directly on the nodes of the inner cylindrical balloon surface [17]. In this study the simulation of stent deployment was carried out with the method described in [17].

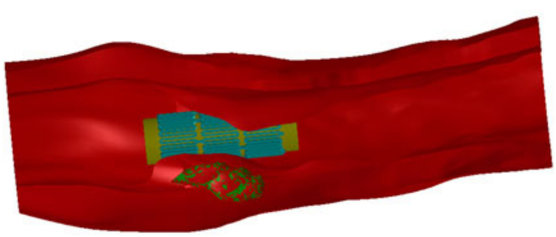

Fig. 1 Geometry of the model: Red, green, blue and yellow colors represent the arterial wall, plaque, stent and balloon geometry, respectively

C. Material Properties

Several components constitute the arterial tissue some of which are collagen and elastin cells. The artery and the plaque can be modeled using several material models. The complexity of the behavior of the arterial tissue can be more accurately described using hyperelastic material models. A Mooney-Rivlin hyperelastic equation was used to define the arterial tissue which depicts the non-linear behavior presented between stress and strain of the arterial tissue and it is defined by a polynomial form [15]. Maurel *et al.* [18] described the strain energy density function, for an isotropic hyperelastic material, in terms of the strain invariants:

$$W(I_1, I_2, I_3) = \sum_{p,q,r=0}^{n} C_{pqr} (I_1 - 3)^p (I_2 - 3)^q (I_3 - 3)^r, \quad (1)$$

where, W is the strain energy density function of hyperelastic material, C_{pqr} is the hyperelastic constants, C_{000}=0, $\lambda_1, \lambda_2, \lambda_3$ are the principal stretches of material and I_1, I_2, I_3 are the strain invariants. The strain invariants are defined in Eqs. (2)-(4):

$$I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2, \quad (2)$$

$$I_2 = \lambda_1^2\lambda_2^2 + \lambda_1^2\lambda_3^2 + \lambda_2^2\lambda_3^2, \quad (3)$$

$$I_3 = \lambda_1^2\lambda_2^2\lambda_3^2. \quad (4)$$

In this study a third order, five parameters Mooney Rivlin model (Eq. (5)) was used:

$$W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) + C_{20}(I_1 - 3)^2 + C_{11}(I_1 - 3)(I_2 - 3) + C_{30}(I_1 - 3)^3, \quad (5)$$

described by Eshghi *et al.* [19] which is a specific form of Eq. (1). The hyperelastic constants of Eq. (5) are given in Table 1. Stress components were derived by differentiating the strain energy function with respect to strain variables.

Table 1 Artery hyperelastic coefficients.

Coefficients	C_{10}	C_{01}	C_{20}	C_{11}	C_{30}
Artery	0.0189	0.00275	0.08572	0.5904	0

The plaque was assumed to be calcified and was modeled as a linear isotropic elastic material with Young modulus 2.7MPa and Poisson's ratio 0.4913. The stent was made of 304 Stainless Steel and modeled as a bi-linear elasto-plastic material with Elastic Modulus 193GPa, Poisson's ratio 0.27 and Tangent modulus 0.692GPa [20]. Regarding the balloon, it was assumed to be made of polyurethane described as a hyperelastic material. The Mooney–Rivlin model was used with C_{10}=1.0318MPa and C_{01}=3.6927 MPa [6].

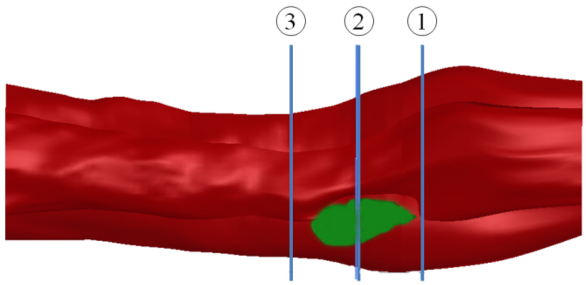

Fig. 2 The model with the selected cross sections

D. Boundary Conditions

In order to perform a steady-state analysis and prevent rigid body motion, certain specific areas must be fixed by applying displacement limitations upon the model's nodes. The artery was fixed at its ends and the stent was tethered in such a way that only radial displacement was allowed. The balloon was in initial contact with the stent and radial displacement (0.8mm) was applied in the inner surface of the balloon. Frictionless contact was assumed for the balloon and the stent contact pair, while artery and plaque were assumed to be bonded. Since the simulation involved large displacements and complex contact analysis, special attention must be taken in parameters like time step, contact algorithm, stiffness and penetration factors.

III. RESULTS AND DISCUSSION

The results are obtained in terms of the stress and the deformation that occur in the model's components. Figure 2 depicts the cross sections where the results are presented. Three different cross sections along longitudinal axis are selected: 1) before the maximum stenotic location, 2) in the maximum stenotic area, and 3) after the stenotic area. The arterial wall deformation and the corresponding von Mises contour maps, of the stent deployment procedure in these cross sections are presented in Figure 3. It is observed that due to the plaque stiffness the artery is deformed more than the plaque, especially in the maximum stenotic area. The stresses caused in the artery wall have a descending ratio when going from the artery's lumen interface to the outer wall surface. The maximum von Mises stresses are observed at the region across the plaque component.

Figure 4 shows the von Mises stress in the arterial wall for a vertical cross section. It is noticed that higher stresses appear in the area behind the stent struts. Figure 5 presents the plaque's von Mises stresses; the maximum plaque stress appears in the contact area between the stent and the artery, in the point where the stent expands and pushes the plaque.

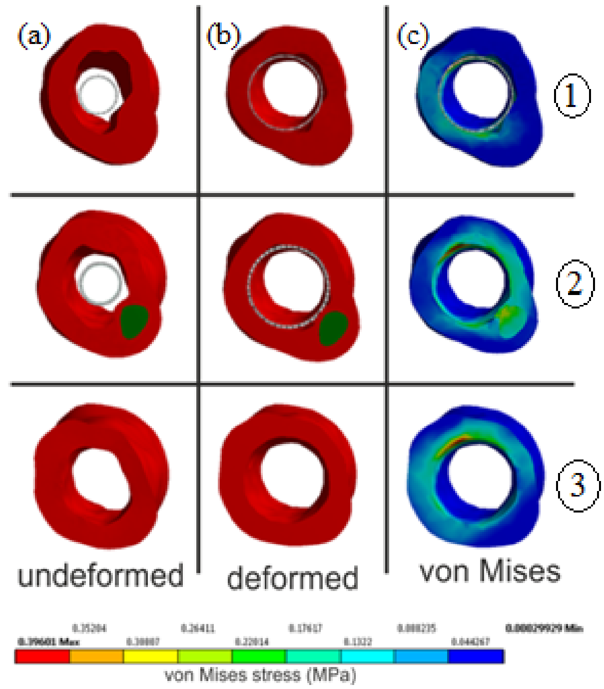

Fig. 3 (a) Undeformed arterial wall and plaque, (b) Deformed arterial wall and plaque, (c) Arterial wall and plaque Von Mises stress, for the three cross sections indicated in Fig.2

Plaque fracture phenomena are most likely to occur in areas associated with high stress [6]. Figure 6 presents the plaque's and artery's volume distribution of von Mises stress. The graph shows that the percentage of plaque volume in the stress range of 0.1-0.15MPa is almost equal to the percentage volume of 0.15MPa-0.2MPa stress range. High von Mises stresses (>0.25MPa) exist in less than 1% of the plaque's stented volume. Regarding the artery it is worth to mention that on average the stresses in the arterial tissue are significantly lower than those in the plaque.

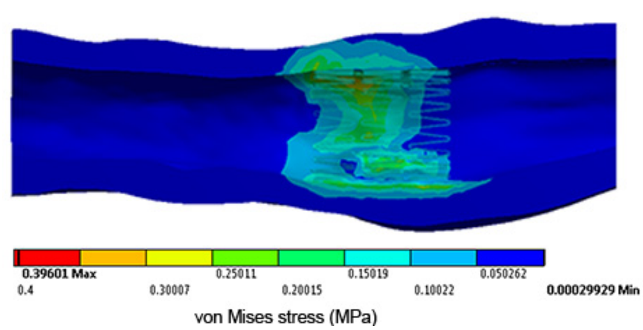

Fig. 4 Arterial wall von Mises stress results

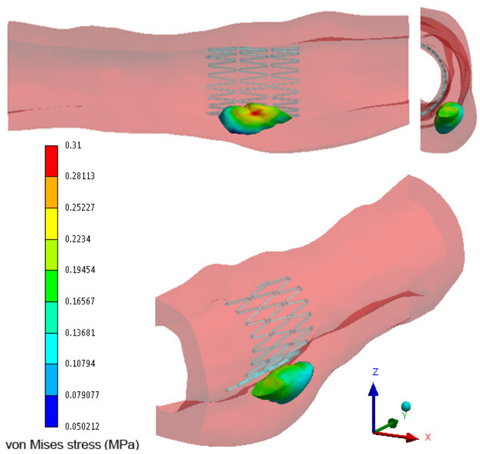

Fig. 5 Von Mises stress results on the plaque component

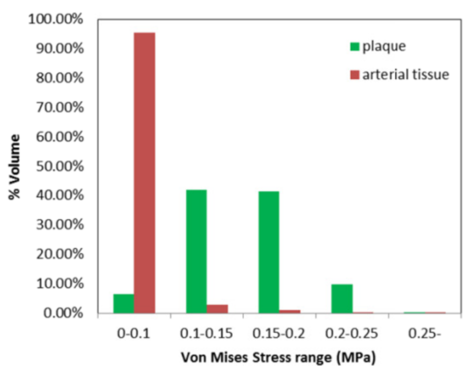

Fig. 6 Von Mises stress percentage volume distribution for the plaque and the arterial wall tissue

IV. Conclusion

In this study the finite element method was used to illustrate the stent implantation and revealed that higher stress occurs in the contact area where the stent pushes the underlying plaque. For the first time, realistic geometries of artery and plaque composition are utilized for simulating stent implantation. For future work, the behavior of different plaque types must be examined, as well as the hemodynamic effect of stent deployment on the wall shear stress distribution which plays a significant role to plaque progression and rupture.

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