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A numerical study on the application of the functionally graded materials in the stent design

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ABSTRACT

Undesirable deformation of the stent can induce a significant amount of injure not only to the blood vessel but also to the plaque. The objective of this study was to reduce/minimize these undesirable deformations by the application of Functionally Graded Materials (FGM). To do this, Finite Element (FE) method was employed to simulate the expansion of a stent and the corresponding displacement of the stenosis plaque. Three hyperelastic plaque types as well as five elastoplastic stents were simulated. Dogboning, foreshortening, maximum stress in the plaque, and the pressure which is needed to fully expand the stent for different stent materials, were acquired. While all FGMs had lower dogboning in comparison to the stents made of the uniform materials, the stent with the lowest heterogeneous index displayed the lowest amount of dogboning. Steel stent showed the lowest foreshortening and fully expansion pressure but the difference was much lower than that the one for dogboning. Therefore, the FGM with the heterogeneous index of 0.5 is expected to exhibit the most suitable results. In addition, the results revealed that the material parameters has crucial effects on the deformation of the stent and, as a result, as a design point of view the FGM parameters can be tailored to achieve the goal of the biomechanical optimization.

1. Introduction

Atherosclerosis is defined as the hardening of a blood vessel due to an atheromatous plaque, which leads to the narrowing or even blocking of the blood pathway inside an vessel [1,2]. Several reasons have been suggested to initiate the atherosclerosis, but after development of the initial plaque, it will undergo the slow process of atherogenesis. There are many hypotheses regarding the atherogenesis, yet we know that in the late atherosclerosis the plaque while become calcified [3]. While the plaque changes, its composition and mechanical properties alter as well [4–6]. Calcification process will generally harden the plaque. One of the main treatments of the atherosclerosis is angioplasty with stenting, where a balloon with a stent will be inserted into the diseased artery. Afterward, when they reached the diseased cite, balloon will inflate, its inflation will expand the stent, then the balloon will be removed. But because of the plastic deformation that occurred in the stent it will remain expanded to function as a scaffold to keep the artery open. The plaque or the artery can be damaged during the process of stent deployment [7]. To prevent this event, an understanding of this mechanical process is deemed needed.

In a Functionally Graded Material (FGM), both the composition and the structure gradually change over the volume, resulting in corresponding alterations in the properties of the material [8]. The general idea of structural gradients first was advanced for composites and polymeric materials in 1972 [8–10]. In 1985, the use of continuous texture control was proposed in order to increase the adhesion strength and minimize the thermal stress in the ceramic coatings and joints being developed for the reusable rocket engine [8]. The developers of this material realized that application of this concept can be extended to impart new properties and functions to any material by gradually changing its texture or composition. Then, these materials were named FGMs. Since then, their applications have been rapidly growing. Nowadays, there are many areas of application for FGMs, and one of the most applicable ones is biomedical application. Biomaterials should simultaneously satisfy many requirements and possess properties, such as nontoxicity, corrosion resistance, thermal conductivity, strength, fatigue durability, biocompatibility, and sometimes esthetics [11]. A single composition with a uniform structure may not satisfy all such requirements. For example, FGMs have been used as dental implants [12], bone tissues [13], intervertebral disc implant [14], prosthetic knees or holes of hip acetabular bone anchor [15,16].

Primary balloon angioplasty has been established as an effective treatment for coronary artery atherosclerosis. However, it has been reported that this method is suffering from a high reocclusion and using

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stent in the balloon angioplasty has proven to improve the results of the procedure [17]. The first use of the stent for cardiovascular purposed was in 1964 [18], but the first coronary stent was implanted in a patient in 1986 [19]. Since then angioplasty with stent has become one of the most common procedures in the world. Due to the mechanical nature of the stenting procedure, mechanical analysis is needed to better understand and improve this device. Finite Element (FE) has proven to be a valuable method for this purpose [20,21]. FEM has been used to investigate the impact of the balloon folding on the stent implantation [22]. Moreover, it has been shown that change of the geometrical parameters can reduce the dogboning and foreshortening effects [23–25]. Furthermore, asymmetric geometry was purposed to reduce the dogboning [26]. Additionally, optimization methods have been used extensively to find an optimum design for the stent [27–30].

Despite all of the studies that have been done on this subject, still undesirable deformations, such as dogboning are increasing the restenosis chances and the stent performance is inadequate. In this study, FE method has been implemented to investigate deployment of a stent with different material properties as the design parameter. Stress in the plaque site, needed pressure to fully expand the stent, dogboning and foreshortening, have all been used to compare the performance of different stent designs, during deployment in an artery affected by atherosclerosis. The main goal of this study was to find a proper heterogeneous index to reduce/minimize the undesirable deformation of the stent.

2. Method

2.1. Geometry of the FE components

A balloon expandable Palmaz-Schatz stent geometry was used as the geometrical model. The length of the stent was 9.86 mm, its unexpanded inner radius was 0.70 mm with the thickness of 0.05 mm. The coronary artery was modeled as a tube with the inner radius, thickness, and length of the 1.8, 0.4635, and 14 mm, respectively. The meshed model of the blood vessel, plaque, and stent is demonstrated in Fig. 1. Second order tetrahedral elements were used to mesh the geometry. The number of elements for each part of the model is listed in Table 1.

2.2. Material properties of the FE components

Elasto-plastic materials were used to model the mechanical behavior of the stent. Two common materials that have been used in stents, namely SS316L steel and cobalt chromium alloy (CoCr L605) [31,32]. Their material properties are summarized in Table 2. These two metals were employed to represent the material properties of the stent. The FGM was a combination of these two materials, where the FGM stent material combination would start as a cobalt chromium at one end,

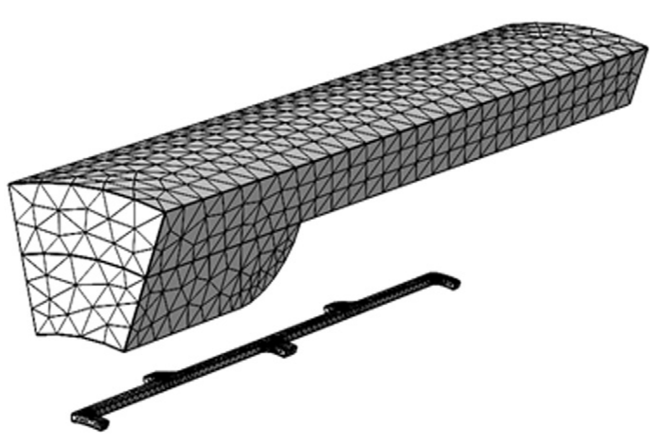

Fig. 1. Meshed symmetric geometry, including the blood vessel, plaque, and stent.

Table 1
Number of elements in each part of the model.

	Artery	Plaque	Stent	Total
Number of elements	3498	1228	3845	8571

gradually changing to complete steel in the middle of the stent, and eventually changing back to CoCr alloy at the other end. The alterations in the material properties of the FGM stent over its length is described by Eq. (1). Where f is the given material properties, D is the distance from the middle, l is the length, and n is the heterogeneous index. Large plastic strain theory was used to account for the large plastic deformation of the stent [33].

$$f(\mathrm{l}) = (f_{out} - f_{in})\left[\frac{\mathrm{D}}{\mathrm{l}}\right]^{\mathrm{n}} + f_{in} \tag{1}$$

It has been shown that hyperelastic material properties can adequately describe the stress-strain behavior of the both coronary artery and plaque [4,34,35]. A hyperelastic material is defined by its elastic Strain Energy Density Function (SEDF), which is a function of the elastic strain. The hyperelastic formulation normally gives a nonlinear relation between stress and strain, as opposed to Hooke's law in linear elasticity [36]. Five parameter Mooney-Rivlin hyperelastic constitutive equations was used to describe the material properties of the coronary artery and cellular plaque [34,35]. Whereas hypocellular and calcified plaques material properties were defined by the Ogden model [4]. Material properties of the hyperelastic bodies are listed in Table 3. Eq. (2) describes the general form of the SEDF in terms of the strain invariants for an isotropic hyperelastic material [36,37].

$$\mathrm{W}(\mathrm{I}_1, \mathrm{I}_2, \mathrm{I}_3) = \sum_{\mathrm{i,j,k}=0}^{\infty} \mathrm{a}_{\mathrm{ijk}}(\mathrm{I}_1-3)^M(\mathrm{I}_2-3)^N(\mathrm{I}_3-3)^0, a_{000} = 0 \tag{2}$$

where I_1, I_2, and I_3 are the strain invariants, W is the SEDF, and $\mathrm{a}_{\mathrm{ijk}}$ are the hyperelastic constants. Coronary artery and stenosis plaques can be assumed to be incompressible [38–40]. An isotropic hyperelastic incompressible material is characterized by a SEDF which is a function of two principal strain invariants [41–43]. These two are defined by:

$$\mathrm{I}_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \tag{3}$$

$$\mathrm{I}_2 = \lambda_1^2\lambda_2^2 + \lambda_1^2\lambda_3^2 + \lambda_2^2\lambda_3^2 \tag{4}$$

Therefore, Mooney-Rivlin hyperelastic model can be used as a suitable SEDF to model an incompressible isotropic hyperelastic material [44,45]. Eq. (5) describes the five parameter Mooney-Rivlin model as a function of strain invariants.

$$\begin{aligned} \mathrm{W} = {} & a_{10}(\mathrm{I}_1-3) + a_{01}(\mathrm{I}_2-3) + a_{20}(\mathrm{I}_1-3)^2 + a_{02}(\mathrm{I}_2-3)^2 \\ & + a_{11}(\mathrm{I}_1-3)(\mathrm{I}_2-3) \end{aligned} \tag{5}$$

Another hyperelastic model that has been used to describe the biological tissues is the Ogden [41,46,47]. Ogden has a polynomial form in terms of the stretch ratios as its variables instead of the invariants

Table 2
Mechanical properties of the homogenous stent materials.

	E (GPa)	ρ (kg/m^3)	ν	S_y (MPa)	E_t (MPa)
Steel	193	7800	0.27	207	692
CoCr alloy	243	9700	0.30	476	680