Dynamic characteristics of composite coronary stents after implantation

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Abstract

The dynamic characteristics of composite coronary stents of poly-l-lactic acid (PLLA) coated magnesium (Mg) alloy were investigated using the finite element method (FEM). Firstly, the effects on dynamic performances of stents were considered, such as stent materials, large deformation of expansion and springback during stent implantation, residual stress after implantation, degradation of PLLA coating, and vascular constraints. Secondly, variations in the dynamic characteristics of the blocked artery after stent implantation were calculated. The natural frequencies and corresponding vibration modes of stents and arteries, as well as the response under harmonic excitation were numerically simulated. The results show that, the natural frequency of the composite stent is much smaller than that of the Mg alloy stent. Each natural frequency of the stent after a large deformation of expansion and springback significantly decreased compared with that of the initial stent. The existence of residual stress has a minor effect on the natural frequencies of the stent and does not change the vibration modes. However, degradation of the PLLA coating and vascular elastic constraint have distinct influences on the frequencies of stents. Modal analysis results indicate that bending, torsional and breathing modes occur in the first five vibration modes. Moreover, there are differences in the natural frequencies and vibration modes among healthy, blocked and stent-implanted arteries. These results are helpful for understanding the dynamic behavior of the vascular system after stent implantation and have guiding significance in stent design.

1. Introduction

Intracoronary stent implantation has been largely used in the treatment of coronary atherosclerosis which is the primary cause of deaths worldwide annually. It was first performed in human by Sigwart and Puel in 1986 [1] and was proved to be effective in reducing acute vessel occlusion and restenosis. This technique is much safer than coronary bypass operation due to its minimally invasive treatment. During operation, the stent is delivered to the coronary artery lesion site by catheter and then expanded by a balloon, so as to support the vascular wall and keep the blood at the lesion site flowing smoothly. The stent currently used in clinic is a mesh shell device made of metal or polymer material. The newly generated stents creatively utilize bioresorbable materials to reduce ischemia-reperfusion injury and stent thrombosis. Current studies for coronary stents focus on the biomechanical related to the complications and design optimization of stents in materials and structures. From the previous works by Schwartz et al [5] and Kornowski et al [6], the occurrence of ISR correlates with the inflammatory reaction induced by vascular injury after coronary stenting. Geith et al [7] verified the severity of softening of coronary
arteries is linked to the strut orientation, indentation pressure, and position, by correlating the mechanical response with structural parameters of the injured tissue. Therefore, design requirements for stents should be taken into account, such as support intensity, fatigue resistance, flexibility and so on. Studies on the mechanical behaviors of stents is essential to resolve the existing problems of stent intervention. Imani et al. [8] investigated the mechanical characteristics of two commercial stents during implantation in a plaque blocked artery by calculating their stress distribution, radial gain, outer diameter changes and dogboning. Sanjay et al. [9] explored the influence of stent shape on hemodynamics. Moreover, some researchers studied the stent fatigue behaviors under periodic pulsatile loading in consideration that it could lead to the fracture of stent struts [10–12].

Finite element analysis (FEA) has been extensively used to simulate various static mechanical behaviors of stents during implantation, such as the influence of the geometry on the stent behavior [13], variation of applied loads on longitudinal stent deformation in a patient specific coronary artery segment [14], as well as the effectiveness of cardiovascular stenting by parametrically analyze the process of crimping and expansion [15]. Recently, it was used to optimize the structural characteristics of stent designs [16, 17]. Indeed, the numerical simulation could greatly reduce experimental costs and be helpful to make clear that are difficult to perform in realistic conditions.

To date, it is still scarce for the investigation to dynamic behaviors of stents in service. Actually, after a stent is implanted into the lesion location, it would vibrate as the artery pulsation in service. The implanted stents, as an accessional elastic element, are bound to change local stiffness in the artery, this will lead to dynamic properties of vessel with implanted stents being different from one of the artery without the stents. Such variation might cause the fatigue generation of stent and turbulence of blood flow leading to potential damage to heart valves. Thus, it is required to analyze the dynamic behaviors of stents, such as the natural frequency and the vibration mode of stents. Earlier, Ma and Shang proposed the investigation on the free vibration of an elastic stent, where the stent was modeled as an orthogonally anisotropic thin shell based on the Flügge shell theory [18]. Kumar et al. estimated the natural frequency of stents with different materials based on the finite element method [19]. Further investigation of dynamic characteristics of stents implanted in arteries should be carried out, such as the effect of residual stress after expansion and springback. The stent has variations in structural geometry and internal stress in the large elastoplastic deformation, so the dynamic performance of the stent might be varied. With that in mind, before investigating the dynamic performance of stents, it is need to analyze the elastoplastic large deformation of stent in the implantation process. Especially, stents are attached to blood vessel walls in servicing process, vascular constraints should be considered in the calculation.

Viewed from another perspective, arteries implanted with stents would change the initial vascular structure and stiffness, which could change the dynamic properties of arteries. Therefore, vibration analysis on three kinds of arteries in healthy, blocked and stenting situations will also be performed.

There are two aims of the present paper. One is to investigate the dynamic characteristics of Stent-C under vascular constraints by FEA technology. The natural frequencies and corresponding vibration modes of stents are essential to analyze the dynamic response of stents. Modal analysis can obtain the natural vibration characteristics of stents, that is, the specific natural frequency and modal shape of each mode. Harmonic response analysis is to determine the steady-state forced response of the structure under harmonic load by means of frequency sweep. Harmonic response analysis could predict the sustained dynamic characteristics of the structure and verify whether the design can overcome resonance, fatigue, and other harmful effects caused by forced vibration. Therefore, modal analysis and harmonic response analysis will be conducted in this work. Obviously, the dynamic performance of the implanted stents differs from that of initial stents, because residual deformation and stress could generate during the stent implantation. Therefore, the dynamic analysis for stents should be based on configuration after expansion and springback, meanwhile, the residual stress should taken into consideration. For comparison, the dynamic analysis will also be conducted on two kinds of single-material stents, namely Mg stent (Stent-M in short) and PLLA stent (Stent-P in short). The situation of bare stents without vascular constraints will be considered to analyze the influence of vascular constraints on dynamic characteristics of stents. The second aim is to analyze the variation of dynamic characteristics of arteries after stents were implanted, by comparing with healthy arteries and those blocked by plaque.

2. Methods

2.1. Geometry and meshing

The 3D numerical models of stents and arteries were constructed by a commercial software Unigraphics NX (Siemens PLM Software). HyperMesh (Altair HyperWorks software) was used to perform finite element meshing.
The structure of stents is based on the Magic stent (Biotronik, Berlin, Germany) [20]. As shown in figure 1(a), the considered three types of stents have the same geometry and size from an exterior view. It is a cylindrical latticed shell with circumferential structure symmetry and periodicity in the longitudinal direction. There are seven wavy rings along the axial arrangement, which act as circumferential support structures. Every two adjacent wavy rings are connected by four straight bars, which can perform connection function. The outer diameter and the thickness of the cylindrical shell are respectively 1.4 mm and 8.9 mm. From the cross-sectional view of a strut in each stent (see figure 1(c)), the three kinds of stents have different morphology. Stent-C is of bi-material coated construction, but Stent-P and Stent-M are single-material construction. The outside dimensions of rectangular cross sections are with a width of 0.08 mm and a height of 0.15 mm. The thickness of outer layer of Stent-C is 0.02 mm. Using Hypermesh software, the stent geometry is meshed into 199624 8-node reduced integration linear brick (C3D8R) elements with an average element characteristic length of 0.02 mm, which was arrived at through a mesh convergence study. A part detail of the mesh is shown in figure 1(b).

According to the description of vascular structure [21], a healthy artery is composed of three layers: intima—the innermost layer consisting of a single layer of endothelial cells, media—composed of smooth muscle cells and a network of elastic and collagen fibrils, adventitia—the outermost layer consisted of thick bundles of helical collagen fibrils. One kind of idealized structure was modeled and analyzed in this study by referring to the ideal symmetrical treatment in the finite element simulation studies of stents [22, 23]. The stent was located in the middle of a length of artery containing plaque. A half section view along axial direction of the model is shown in figure 2. The artery of healthy region was modeled as a cylindrical tube with a total length of 18 mm, an outer diameter of 4.58 mm. After calculating several models with different vascular length, 18 mm is the final determined size. Compared with 18 mm, the structural size over this length do not cause any difference in the mechanical behavior of the intermediate segment structure. That is, there is no edge effect. The three layers have thicknesses of intima (0.24 mm), media (0.32 mm) and adventitia (0.34 mm) according to the data obtained by Holzapfel et al (2005) [24]. The plaque structure is assumed to be ideally axisymmetric with a length of 5 mm, and its thickness gradually increases from zero at the distal section to 0.5 mm at the middle section. The overall
structure of artery layers and the plaque were meshed into 41748 8-node linear brick, reduced integration, hourglass control (C3D8RH) elements, with an average element characteristic length of 0.2 mm. The stent in this model has the same structure and meshing elements as the above-introduced bare stent.

2.2. Material constitutive model

The material type of Mg alloy is AZ31l, its parameters are derived from a tensile experimental data at 37° [25]. Ramberg-Osgood elastoplastic model is employed to approximate the experimental data:

\[ \varepsilon = \frac{\sigma}{E} + K(\sigma/E)^n \]

where \( \sigma, \varepsilon, E = 43.5 \) GPa, \( K = 3.5 \times 10^{11} \) and \( n = 5.82 \) [26] are in turn the true stress, the true strain the elastic modulus, the hardening coefficient and the hardening index of the material. The density and the Poisson’s ratio of Mg alloy is \( 1.77 \) g cm\(^{-3}\) and 0.35 respectively.

PLLA material adopted the ideal elastic-plastic model with an experimentally determined Young’s modulus of \( E = 3.3 \) GPa and a Poisson’s ratio of \( \nu = 0.3 \) based on experimental testing at 37 °C by Nasim Paryab et al [27]. The true stress \( \sigma \) increases linearly with the increase of true strain \( \varepsilon \) when \( \varepsilon \) was less than the yield strain \( \varepsilon_y \), and \( \sigma \) equals with the yield stress \( \sigma_y \) of 51.5 MPa when \( \varepsilon \) is greater than \( \varepsilon_y \). The density of PLLA is 1.25 g cm\(^{-3}\).

The material properties of the three artery layers and plaque all perform hyperelastic and nearly incompressible in mechanical behaviors. Considering that the study is focused on the deformation and vibration of stents rather than the mechanical behavior of artery, it is acceptable using an isotropic constitutive model for artery although an anisotropic description for artery is more appropriate physically. The intima, media and adventitia of artery were all described by the third-order Ogden hyperelastic strain energy potential model based on the data from Zahedmanesh et al [28]. The plaque is modeled using a similar first-order Ogden hyperplastic model based on experimental plaque testing by Loree et al [29]. This hyperelastic model was obtained by fitting to the experimental stress–strain data, its energy function is given as follows:

\[ W = \sum_{i=1}^{N} \frac{2\mu_i (L_i^{\alpha_i} + L_i^{\beta_i} + L_i^{\gamma_i} - 3)}{\alpha_i} + \sum_{i=1}^{N} \frac{1}{D_i (J - 1)^{2i}} \]

where \( N \) represents the order of the hyperelastic model, it takes 3 for the three artery layers and 1 for the plaque, \( L_i \) (i = 1, 2, 3) denotes deviatoric stretches in three principal directions, \( \mu_i \) (MPa), \( \alpha_i \) and \( D_i \) are dimensionless material parameters, \( J \) is the elastic volume strain. These parameters take different values for the vascular layers and plaque. A Poisson’s ratio of 0.49 was specified for the nearly incompressibility of vascular layers and plaque, so the value given to \( D_i \) was infinitesimal, and to \( D_2 \) and \( D_3 \) was 0. Parameter values of the Ogden hyperplastic model used in this work for three arterial layers and the plaque referenced to the data from Schiavone et al [30].

Considering that the stiffness of the vessel is much smaller than the stent, the modal calculation of stents is easily interfered by vascular modes. In the part of calculating the influence of vascular constraints on vibration characteristics of the stent, the density of three artery layers and the plaque were set as zero in finite element analysis. In this way, the vessel does not vibrate itself but provides boundary constraints, and only the vibration modes of the stent can be calculated. However, in the part of calculating the vibration characteristics for the whole system of the artery implanted with a stent, the densities of three vascular layers and the plaque were set as 1.07 g cm\(^{-3}\) and 1.45 g cm\(^{-3}\) respectively.

2.3. Finite element simulations

All finite element analysis were performed using Abaqus Standard (Simulia, Providence, RI, USA) implicit solver. Dynamic analysis on considered stents and arteries will be carried out.

The interface between adjacent arterial layers and between two materials of Stent-C is assumed to be connected properly, that is, there is continuity of normal and tangential displacements and stresses. Hence, ‘Tie’ command of Abaqus is used in each pair of adjacent surfaces. For the structure of a stent implanted in a coronary artery, contact between the stent and the artery was defined as hard contact for normal contact behavior and a friction coefficient of 0.2 for the tangential contact property. Considering that the coronary arteries are not embedded in the myocardium, so the boundary conditions can be set as free and unconstrained. The axial and circumferential displacements at both ends of the vessel are constrained, but they are free in the radial direction, which conforms to the real state of vasoconstriction and relaxation.

Real cases, the stent is expanded by the balloon under pressure. According to the experimental observation and literature investigation [31], after the stent is fully expanded, it’s basically a uniform cylindrical shell. This is consistent with the final structure obtained by applying displacement loads. In addition, since the deformation of the stent is stable in the process of expansion, there is a one-to-one correspondence between pressure and displacement. This paper focuses on the dynamic characteristics of the stent after it is fully extended, and the deformation structure during expansion process does not affect the results. Therefore, in order to facilitate
calculation, it is reasonable to replace the pressure load with displacement load. In some researches, the method of loading displacement was also used for simulation calculation of stents \([32]\).

For the situation of a stent implanted in an artery, static analysis for expansion and springback should be performed before the dynamic analysis. There are three static analysis steps: firstly a smaller radial displacement of 0.2 mm was applied on the inner surface of stents to smoothly build a contact relation between the stent and the plaque (contact); secondly the displacement load was modified to 0.8 mm to make the stent inner diameter reach 3.2 mm (expansion); thirdly the load was removed followed by elastic recoil of the stent and artery (springback). After that, aiming at the deformed structure of stents with residual stress, dynamic analysis was performed: the first step was to execute frequency calculation of linear perturbation, the second step employed the steady-state dynamics based on modals to calculate response under a harmonic load with 1Mpa amplitude. For initial stents and arteries before stent implantation, dynamic analysis can be performed directly.

When simulating the degradation of PLLA layer, it is performed after expansion and springback of entire stent. After the entire stent expanded and recoiled, the residual deformation and stress generate. The next step is to remove PLLA outer layer. In the implementation of finite element analysis, we employed ‘deactivate’ function in ABAQUS order ‘model change’. In this way, we are able to implement removal of PLLA outer layer part of the model by deactivating elements after the steps of springback.

3. Results and discussions

As natural frequencies and vibration modes are the essential features of dynamic behaviors of structures, we calculated the first five order natural frequencies and corresponding vibration modes of the structures studied in this work, as well as the displacement response of the structure to the frequency at several frequencies. In section 3, several influences on the vibration properties of stents were discussed, including different material composition of stents, deformation process of expansion and springback, degradation of PLLA coating, as well as vascular constraints. In section 3.2, the differences of dynamic behaviors among healthy, blocked and stent implanted arteries were compared.

The deformed structure of Stent-C implanted in a coronary artery after expansion and springback is shown in figure 3.

Figure 3. Deformed shape of Stent-C implanted in a coronary artery after expansion and springback.

3.1. Dynamic characteristics of stents

3.1.1. Natural frequencies

Considering that the research focus on the dynamic characteristics of stents, so the mesh convergence study was done for the composite stent (Stent-C) model. Five different quantity of elements are divided for the stent structure, which takes the value of 49768, 80536, 199624, 357432 and 1446072 respectively. The natural frequencies of Stent-C in each case are calculated. The convergence of the first order natural frequency of Stent-C is selected as the evaluation standard, and the convergence analysis results are shown in figure 4. The result turns out it is reasonable for the stent being meshed into199624 elements.

The first five natural frequencies of considered stents are listed in table 1. It shows that the natural frequency of Stent-C is slightly larger than that of Stent-P and much smaller than that of Stent-M, regardless of the initial or the implanted stents. Although the considered three stents have the same geometric structure, the free vibration characteristics are different because their materials and internal residual stresses are all different.

In contrast to the initial stents, the structural geometric dimension of an implanted stent changes after expansion and springback due to the elastoplastic large deformation, which will affect the characteristics of free vibration of the stent. Although the stents after expansion and springback remain similar geometric appearance to the initial stents, their overall structures can enlarge. Especially, the elastoplastic large deformation during the
implantation can generate residual stresses inside the stents. These factors would bring the difference of natural frequency between the implanted and the initial stents.

As seen from the results in table 1, compared with the initial stents, each order natural frequency is reduced for the three implanted stents. For example, the first-order natural frequency (fundamental frequency) for Stent-C, Stent-M and Stent-P are in turn reduced by 8.3%, 10.6% and 8.7%.

It should be noted that the first two order frequencies for initial Stent-M and Stent-P are quite close. However, the variation of the first two order frequencies for the two implanted stents differs respectively 5.4% and 4.1%. Moreover, the relative difference between the first and second order frequencies for the initial Stent-C is only 1.8%, while that of the implanted Stent-C becomes 2.8% and the variation is small.

To examine the influence of the degradation of the PLLA outer layer on the natural frequencies of Stent-C, table 1 also lists the natural frequency of the stent in the case that the thickness of the PLLA layer of the implanted Stent-C is reduced to zero. After PLLA layer is entirely degraded, the fundamental frequency increases by about 5%, while the increases of the other four frequencies are about 4% ~ 9%.

Generally, the existence of residual stress may more or less affect the vibration characteristics of structures. For the degraded implanted and non-degraded implanted stent, the natural frequencies are also calculated in the case of not taking the residual stress into account, which is listed in table 1. Compared to the corresponding results with residual stress, for the non-degraded implanted stent, the fundamental frequency only increased by about 0.2%, and the increase or decrease of the other four frequencies do not exceed 0.9%. For the stent with the PLLA layer entirely degraded, the fundamental frequency is only reduced by about 0.1%, and the relative increase or decrease of the other four frequencies do not exceed 0.8%.

For stents under vascular constraints, the stent was expanded and recoiled in a part of artery with a plaque, thus the deformation and vibration will be influenced by vascular constraints. From figure 3, it can be seen that the artery blocked by a plaque was significantly larger in diameter after stent implantation compared with its initial shape.

### Table 1. First five natural frequencies of the considered stents f (Hz).

| Mode                                      | 1st   | 2nd   | 3rd   | 4th   | 5th   |
|-------------------------------------------|-------|-------|-------|-------|-------|
| Bare Stent-P Initial                      | 2550  | 2560  | 4186  | 4838  | 5806  |
| Implanted                                 | 2327  | 2422  | 3774  | 4420  | 4577  |
| Bare Stent-M Initial                      | 7774.5| 7774.9| 12741 | 14699 | 17595 |
| Implanted                                 | 6951  | 7323  | 11380 | 13298 | 13837 |
| Bare Stent-C Initial                      | 3349  | 3404  | 5672  | 6325  | 7647  |
| Implanted                                 | 3071  | 3156  | 5006  | 5960  | 6012  |
| Residual stress free                      | 3076  | 3130  | 5016  | 5911  | 6067  |
| Bare Stent-C after implantation and degradation With residual stress | 3226  | 3278  | 5259  | 6199  | 6534  |
| Residual stress free                      | 3222  | 3250  | 5261  | 6149  | 6587  |
| Stent-C constrained by vascular wall Implanted | 4891  | 5149  | 6158  | 6263  | 6687  |
From table 1, there are variations in natural frequencies between constrained Stent-C and bare Stent-C. Compared with implanted bare Stent-C, the first five frequencies increase 59%, 63%, 23%, 5% and 11% in turn. Compared with initial bare Stent-C, the first three order frequencies of implanted constrained Stent-C increase 46%, 51% and 9% in turn, while the next two order frequencies decrease 1% and 13% separately.

3.1.2. Vibration modes
The modal analysis is foundational to investigate the dynamic behavior of structures. The results of modal analysis by finite element indicate that the vibration modes corresponding to the first five natural frequencies are roughly similar for the three implanted stents. Figure 5 exhibit the vibration mode shapes of Stent-C in three situations of initial, implanted without vascular constraints and implanted under vascular constraints in the lateral view and the cross-sectional view. Shadows are added in the lateral view so that the outline of the modal shape could be clearly seen.

![Figure 5. First five vibrational modes for initial bare Stent-C, implanted bare Stent-C without vascular constraints and implanted Stent-C under vascular constraints.](image)

It can be seen from figure 5 that the first five vibration modes of initial bare Stent-C can be classified into three categories: (1) Torsional modes: The first-order mode is torsional around the central symmetry axis of the stent, and is accompanied by the radial expansion of the stent, in which the radius of the cross-section gradually increases from the middle to the two ends of the stent along the axial direction. For this torsional modes the half wave numbers $m = 1$ and $n = 0$. In addition, the fourth-order mode can be regarded as torsional modes with half wave numbers $m = 2$ and $n = 0$. (2) Bending modes: The second-order mode is the bending one in the longitudinal symmetry plane, in which any two cross-sections of the stent are not paralleled and the radius of the stents remains unchanged. Like the bending mode of a thin cylindrical shell, the axial half wave numbers $m = 1$ and the circumferential one $n = 1$. The third-order mode is bending one with $m = 2$ and $n = 1$. (3) Breathing modes: For the first four modes, the cross-section of the stent are all circular, however, the fifth-order mode shape is an ellipse from the cross sectional view, where the flattening of the ellipse has a change from one end to the other along the axial direction.
Different from the initial stent, for the implanted bare Stent-C, the first and the third order modes are bending, while the second and the fourth order modes are torsional. The fifth order is still breathing mode, however from cross-sectional view, the ellipse becomes more and more flat from both ends to the middle of stent along the axial direction.

For implanted Stent-C under vascular constraints, the vibration modes are obviously different from that for the implanted bare Stent-C. The first and the second modes are torsional and bending modes respectively, while the third to the fifth modes are all breathing modes exhibiting different forms.

3.1.3. Harmonic response
Harmonic response analysis was performed to obtain the natural frequencies of stents and displacement response value of the Stent-C under sinusoidal excitation. On the one hand, it can verify the natural frequency results obtained by modal analysis, on the other hand, it gives the displacement response value of the stent under a certain load (1 MPa) and the frequency range of structural vibration.

Two situations of Stent-C without and with considering vascular constraints were compared to analyze the influence of vascular constraints on the displacement responses of stents. A node located at the end of two kinds of implanted Stent-C was selected to project their curves of displacement amplitude versus frequency, as shown in figures 6 and 7. The results show that the displacement amplitudes are close to infinitely great in their corresponding natural frequency, which conforms to the resonance characteristics of structures. The resonant frequencies obtained in the amplitude-frequency curves are also the natural frequency of the stent, which are consistent with the natural frequencies results obtained by modal analysis. All the frequency bandwidths where occurred significant increase of displacement amplitude are about 1000 Hz. For implanted bare Stent-C without vascular constraints, the generated displacement amplitude decreases gradually with the increase of excitation frequency. When vascular constraints were considered for implanted Stent-C, the displacement amplitude corresponding to the excitation frequency near the first natural frequency is much larger than that of the next four orders.

3.2. Dynamic characteristics of coronary arteries
The dynamic characteristics of a healthy coronary artery could change if it is blocked by a plaque. Similarly, dynamic characteristics of the blocked artery could change when it was implanted with a stent. The natural frequencies and vibration modes of arteries in the situation of being healthy, plaque blocked and implanted with a stent are given respectively in table 2 and figure 8.

From table 2, when a healthy artery was blocked by a plaque, the first, second and fourth natural frequency decreased 6.73%, 3.45% and 6.94% in turn, while the third and fifth one increased 2.24% and 9.20% respectively. When a stent was implanted in the blocked artery, comparing to the healthy artery, the first, third and fifth natural frequency increased 2.88%, 6.73% and 25.46% in turn, while the second and fourth one decreased 12.07% and 4.42% respectively.

From figure 8, the first four orders vibration modes of considered arteries are similar in the three situations. The first and third vibration modes are bending modes, the second vibration modes are torsional modes, and the fourth vibration modes are trumpet-shaped breathing modes. Different from the healthy artery, the fifth
vibration modes of blocked artery and stent-implanted artery transform into bending modes. It is worth noting that the deformation of the artery part where stent located are relatively small.

### 4. Discussions

This work calculated the free vibration characteristics of a composite stent by using the finite element method, which is rarely reported in the existing researches. From the results of the first five order natural frequencies and vibration modes for considered stents, the influences of materials, degradation of the PLLA coating, residual stress and deformation after implantation and vascular constraints on dynamic characteristics of stents were obtained.

Generally speaking, the greater the natural frequency if either the greater the stiffness or the smaller the mass of an elastic structure. For the initial stents, the difference of mass density between Mg alloy and PLLA is small, but the elastic modulus of the former is an order of magnitude higher than that of the latter. Thus, the higher the proportion of Mg alloy in a Stent- C is, the greater the equivalent stiffness is, and the higher the natural frequency becomes. For the implanted stents, although the natural frequency could be related to the value and distribution of the internal residual stress, the main factor affecting the natural frequency is still the proportion of Mg alloy from the results in table 1.

The natural frequencies of the three stents are reduced after expansion and springback. It is because that the effective stiffness of the implanted stents decreases. In fact, the large deformation in the implantation makes the radius of stents increases nearly twice, and also the curvature of the support beams increases, so that the net structure of the stents formed by support beams and connecting rods becomes sparser. By comparing the natural frequencies of implanted stents in the case of retaining residual stress and the case of removing residual stress, it can be seen that the influence of residual stress on the natural frequency of the stent is slight and can be ignored.

When the stent was constrained by vascular wall, the first five natural frequencies of the stent are greater than that for the stent was free (namely without vascular constraints). Besides, the third to fifth vibration modes are all breathing modes, which indicates that radial breathing deformation is easier to occur than the situation of stent unrestrained by blood vessel.

It is worth noting that, the first-order mode of the bare stent transforms from the initial torsional to the bending mode after implantation, while that of the implanted stent under vascular constraints is torsional mode.
Such variation of modes might be related to the geometry change of the stent after implantation. The initial stent is of larger flexibility in circumferential direction, so it is more prone to torsion. For the implanted bare stent that has undergone the process of expansion and springback, the corrugated rings are greatly stretched out in the circumferential direction due to the expansion, thus the curvature of Ω-shape support beam is reduced. Consequently, the bending flexibility of the stent after implantation becomes greater, and the first mode is bending rather than torsion. When the implanted stent was constrained by vascular wall, the bending stiffness would increase more than torsional stiffness.

Usually, other conditions being equal, the smaller the equivalent stiffness (the larger the equivalent flexibility) of a structure against a kind of deformation, the more likely such deformation is to occur. Results of the above modal analysis confirm that for the considered stents the bending and torsional deformation is much easier to generate, while the radial breathing deformation is not. This just meets the design requirement of stents. Namely, the stent should have a stronger ability to resist deformation in the radial direction, but it needs to have sufficient flexibility of bending as a whole.

Moreover, there is a slight increase or decrease in natural frequencies of the artery after stent implantation. Although the first four vibration modes of arteries are similar, the deformation of the part of artery where connected with a stent are smaller than that of healthy artery, such variation may has in influence on the blood flow.

5. Conclusion

The natural frequencies of the three bare stents after the implantation are all reduced by about 10%. The natural frequencies of Stent-C are smaller than that of Stent-M due to the wrapping of PLLA, moreover, they could increase when the PLLA outer layer is entirely degraded. The existence of residual stress has quite a little effect on the natural frequency of the stent and does not change the vibration mode of the stents. The first four modes of the implanted stents are bending or torsion, and the breathing mode appears in the higher order mode. Vascular constraints increase the first four natural frequencies of the stent but do not change the corresponding vibration modes. On the other hand, after the implantation of stents, the dynamic characteristics of artery would differ.
from the healthy situation. These results may be of guiding significance to analyze the variation of system performance in vivo after stent implantation and the optimal design of stent.

Data availability statement

All data that support the findings of this study are included within the article (and any supplementary files).

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