STUDY ON FLEXIBILITY OF INTRACRANIAL VASCULAR STENTS BASED ON THE FINITE ELEMENT METHOD

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ABSTRACT

Flexibility is a particularly important biomechanical property for intracranial vascular stents. To study the flexibility of stent, the following work was carried out by using the finite element method: Four mechanical models were adopted to simulate the bending deformation of stents, and comparative studies were conducted about the distinction between cantilever beam and simply supported beam, as well as the distinction between moment-loading method and displacement-loading method. A complete process as implanting a stent including compressing, expanding and bending was also simulated, for analyzing the effects of compressing and expanding deformation on stent flexibility. At the same time, the effects of the arrangement and the number of bridges on stent flexibility were researched. The results show that: 1. A same flexibility index was obtained from cantilever beam model and simply supported beam model; displacement-loading method is better than moment-loading for simulating the bending deformation of stents. 2. The flexibility of stent with compressing and expanding deformation is lower than that in the initial form. 3. Crossly arranging the neighboring bridges in axial direction, can effectively improve the stent flexibility and reduce the flexibility difference in various bending directions; the bridge number, has proportional non-linear correlation with the stent rigidity as well as the maximum moment required for bending the stent.

Keywords: Finite element, Intracranial vascular stents, Flexibility, Bridges.

1. INTRODUCTION

The anatomical structure of intracranial vessels is different from that of peripheral vessels and coronary vessels, intracranial vessels are lack of support from surrounding tissues and have many physiological bending, therefore, they have higher demand for flexibility [1, 2]. Flexibility is an important biomechanical property index for intracranial vascular stents [3]. Flexibility refers to the ability of stent to bend, which can be characterized by the reciprocal of the stent rigidity. Poorly flexible stents in curved vessels will cause the vessel straightening and make significant inflexion on the vessel at the end of the stent, which can induce adverse vascular events and restenosis.

To measure the flexibility of stent, the researchers

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carried out vast experimental work. Ormiston *et al*. [4] tested the flexibility of 13 commercial stents in three states by means of a three-point bending method: the original stent, the stent compressed on the balloon and the expanded stent. The experimental results showed that stent design has a great effect on the flexibility, stents of different structure have significant differences in flexibility. The flexibility of compressed stent is mainly determined by the balloon; the flexibility of expanded stent is lower than that of the original stent. Mori *et al*. [5] carried out flexibility test on stents by using a four-point bending method. The results showed that the flexibility of stent structured by supporting rings and connecting bridges is mainly determined by the bridges; studies also found that the stents with centro-symmetric bridges have better flexibility.

Stent is tiny and has complicated processing technology, it costs high that testing mechanical properties by using real stents. In recent years, the finite element method has been widely used in biomechanical property test of vascular stent [6-9]. Wu *et al*. [10] bended the stent by applying moment load at both ends of the stent through the MPC restraining unit; Petrini *et al*. [11] compared the flexibility of two commercially available stents by applying angular displacement load at both ends of the stent. M. Azoouzi *et al*. [9], fixed one end of the stent and applied moment load at the other end of the stent to test the flexibility. When we talked about implanting stent into target vessels, which actually includes the compressing deformation, the implanting process and the expansion of the stent in the vessels. Among above research work, only Petrini *et al*. expanded the stent, and found that flexibility of the expanded stent was lower than that of the original stent; but the compressing deformation of the stent was ignored in simulation. F. Auricchio *et al*. [12] reconstructed the human carotid artery model by using computed tomography angiography (CTA) and bended the compressed stent along the vascular axis and then expanded it in the vascular model, they indirectly evaluated the flexibility of the stent through analyzing the stress and straightening effect made by stent on vessel wall. Ragkousis *et al*. [13] reconstructed the coronary artery model, expanded the stents and calculated the relevant mechanical properties. The stent properties evaluated in the reconstructed vascular model are more in line with the actual situation, but it is computational cost and the results are limited to the established vascular models.

The studies on optimizing flexibility are mainly concentrated in the effects of the structures as well as the sizes of supporting rings and bridges. Matthieu *et al*. [14] found that reducing the circumferential number of supporting rings could increase the supporting rigidity and flexibility of the stent, but at the same time increase the axial shorting. Meanwhile, some studies show that the longer the length of the support ring, the easier the expansion of the stent, but its ability to adhere and across curved vessels is worse.

Bridges could be divided into the flexible ones and the rigid ones, flexibly connected stents had better flexibility and could compensate for the axial shorting during the expansion process [15]. Studies of Xiang Shen *et al*. [16] showed that the stent with S-type bridges had the best flexibility, and the stent with L-type bridges had the worst flexibility; the increase of width and thickness of the bridges would reduce the flexibility of the stents; the increase of length of the connecting rods could improve the flexibility; the width of support rings and curvature radius of wave rings had less effects on the flexibility. Liqing Zheng *et al*. [17] found that the width of stent bridges had greater impact on the flexibility while the thickness of stent wall has less effect. M. Azoouzi *et al*. [9] compared the effects of four bridges, denoted as V-type bridge, N-type bridge, asymmetric V-type bridge and asymmetric N-type bridge on the flexibility of stent, and the results showed that the asymmetric N-type bridges had the best flexibility. Ragkousis *et al*. [13] found that the self-contact of stent would increase the rigidity when the stent was bent; and the connection between adjacent support rings had an important effect on stent flexibility. Bae *et al*. [18] increased the length of the bridges and stretched the bridges into the interior of the support rings to increase stent flexibility.

Different mechanical models were undertaken to simulate the bending deformation of stents [9-11, 19], however, there were no reports about comparison of them. To explore an effective model for testing stent flexibility, in this paper, four typical mechanic models were analyzed and compared by using the finite element method.

Accumulation of the plastic strain is a typical failure mode [20-22], but the influences of compressing and expanding deformation were neglected in most researches about the stents' flexibility. Therefore, the effects of compressing and expanding deformation on stent flexibility were studied in this paper.

All the reported literatures optimized the stent's flexibility by changing the structure of the stent bridges, herein, a new method was explored to optimize the stent's flexibility by arranging the bridges crossly.

2. MATERIALS AND METHODS

2.1 Mechanical Models of Bending Deformation

The bending deformation ability of a beam is usually evaluated by bending rigidity EI. E is the elastic modulus of the material and I is the modulus of bending section. The bending deformation of stent can also be regarded as a mechanic problem of beam bending deformation. The cross-sectional shape of the stent is complex, and the cross-sectional shapes in various axial location are quite different, as a result, the modulus of bending section of the stent can't be directly calculated. But by Eq. (1) the bending rigidity of the stent can be easily calculated.

$$EI = M(x) \times \rho(x) \tag{1}$$

$M(x)$ refers to the moment applied on the cross section. $\rho(x)$ refers to the curvature of the beam at the cross section.

Based on the two basic forms of cantilever beam and

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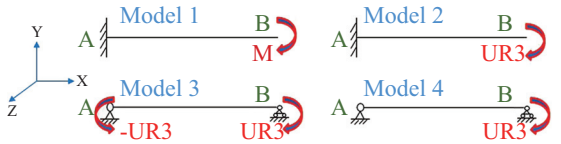


Fig. 1 Four force diagrams for stents bending deformation: Model 1 cantilever beam with moment load; Model 2 cantilever beam with angle displacement load; Model 3 simply supported beam with angular displacement load at the both ends; Model 4 simply supported beam with angular displacement load at the right end.

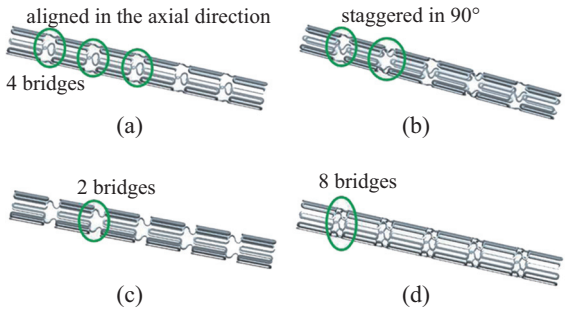


Fig. 2 3D models of stents: (a) noncross-4 bridge stent, (b) cross-4 bridge stent, (c) 2 bridge stent, and (d) 8-bridge stent.

simply supported beam, four mechanical models were given for the bending simulation and flexibility analysis of the stent.

Model 1: the left end (A) of the stent was completely fixed and bending moment was applied on the right end (B). Model 2: Point A of the stent was fully fixed, and an angular displacement load was applied on point B. Model 3: the movements of point A to X, Y, and Z directions and the rotations around the X and Y axes were restricted, and the angular displacement around the Z axis was applied; the movement of point B to Y axis was restricted, and the angular displacement around the Z axis was applied. Model 4: the movements of point A to X, Y, and Z directions and the rotations around the X and Y axes were restricted; the movement of point B to Y axis was restricted, and an angular displacement around the Z axis was applied. The force diagrams of four models are shown in Fig. 1.

Based on the angular equation of the beam Eq. (2), the rigidity computational formula of models are as Eqs. (3-5).

$$\theta(x) = y' = \int \frac{M(x)}{EI} dx + C \tag{2}$$

$\theta(x)$ represents the rotation of the beam.

The computational formula of the rigidity for Model 1 and Model 2 is:

$$EI = \frac{M \times L}{\theta_B} \tag{3}$$

L represents the length of the beam (i.e. the length of the stent).

The computational formula of the rigidity for Model 3 is:

$$EI = \frac{M \times L}{2 \times \theta_B} \tag{4}$$

The computational formula of the rigidity for Model 4 is:

$$EI = \frac{M \times L}{6 \times \theta_A} = \frac{M \times L}{3 \times \theta_B} \tag{5}$$

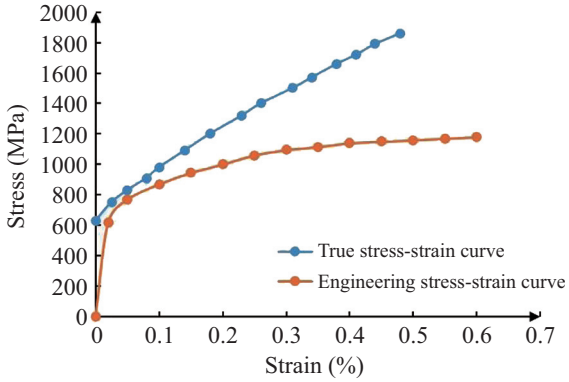


Fig. 3 Stress and strain relationship of L605 CoCr alloy.

2.2 Finite Element Simulation

2.2.1 Geometric Models of Stents

Four stents were designed using the 3D modeling software Pro/E, denoted as noncross-4 bridge stent, cross-4 bridge stent, 2-bridge stent and 8-bridge stent, as shown in Fig. 2. The stents had a typical structure as supporting rings connected with bridges. The adjacent support rings of the noncross-4 bridge stent were connected by four S-type bridges, each set of bridges was aligned in the axial direction. The cross-4 bridge stent was obtained on the basis of the noncross-4 bridge stent by re-arranging the bridges, the neighboring bridges were staggered in 90°. The cross-4 bridge stent was used to study the effect of arrangements of bridges on stent flexibility. The 2 bridge stent and 8-bridge stent were used to analyze the effect of bridge number on stent flexibility.

As the stent is periodic in the axial direction, in order to reduce the computational cost, the model established in this paper has only six groups of supporting rings.

2.2.2 Model of Stent Material

The stent material is L605 CoCr alloy, isotropic and Von Mises yield criteria are used to describe the elastic-plastic property of the stents, elastic modulus is 240 GPa, poisson's ratio is 0.3, yield strength is 630 MPa, the ultimate strength is 1165 Mpa, extending rate is 55%. True stress-strain curve is shown as in Fig. 3.

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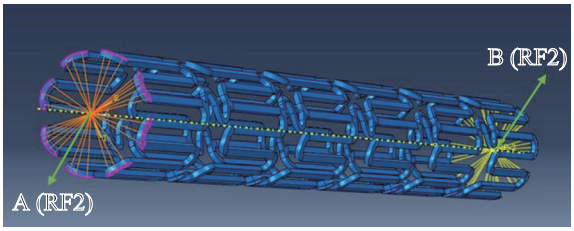


Fig. 4 Coupling both ends of the stent with control points for uniformly loading.

Table 1 Boundary and load definition for Set A.

Model	Boundary Condition		Load
	A	B	
Model 1	U1 = U2 = U3 = UR1 = UR2 = UR3 = 0	Freely	Moment Load: CM3 = 2 N/mm
Model 2	U1 = U2 = U3 = UR1 = UR2 = UR3 = 0	UR3 = 1.5 rad	(Rotational angle displacement)
Model 3	U1 = U2 = U3 = UR1 = UR2 = 0; UR3 = -1	U2 = 0; UR3 = 1 rad	(Rotational angle displacement)
Model 4	U1 = U2 = U3 = UR1 = UR2 = 0;	U2 = 0; UR3 = 1.5 rad	(Rotational angle displacement)

Table 2 Steps and load definitions for Set B.

Steps	Load
Step 1: Compressing	Displacement load U1 = -0.2 mm
Step 2: Recoiling	
Step 3: Expanding	Displacement load U1 = 0.5 mm
Step 4: Recoiling	
Step 5: Bending	Rotational angle displacement on point B UR3 = 1.5 rad

2.2.3 Boundary Conditions and Load Definition

The simulation in this paper includes three parts: Set A, the comparison study of mechanical models of stent bending deformation; Set B, the effect of the compressing and expanding deformation on flexibility; Set C, the effects of arrangement and number of bridges on flexibility. In order to uniformly apply the load on the stent, taking the compression and expansion of the stent into account at the same time, two control points A (RF1), B (RF2) at the intersection of the both end faces of the stent and the axes were set in the interaction module, and the supporting rings at both ends were coupled with A and B, respectively, as shown in Fig. 4.

Set A: The bending simulation of noncross-4 bridge stent was carried out according to the four mechanical models as shown in Fig. 1, and the boundary and load condition of four models are shown in Table 1.

Set B: First, radial compressing and recoiling were acted on the noncross-4 bridge stent, and then expanding and recoiling were carried out, and finally the stent was bent by using the aforementioned way of Model 2. Specific steps and load definition are shown in Table 2.

Set C: The stents are periodically repeatable along the circumferential direction, and in one period, the stent may have different rigidities in various circumferential directions. Therefore, with the local coordinate systems, three tests were carried out on Cross-4 bridge stent at regular intervals in the 45° range, namely Cross-a, Cross-b, Cross-c. In the model Cross-a, bending load was applied on the stent around the Z axis in the local coordinate system CYSY1, as shown in Fig. 5; the coordinate system CYSY1 was rotated by 22.5° and 45° around the Y axis to produce local coordinate system CYSY2, CYSY3 for Cross-b, Cross-c. In the same way, four tests were carried out on Noncross-4 bridge stent at regular intervals in the 90° range, recorded as Noncross-a, Noncross-b, Noncross-c and Noncross-d.

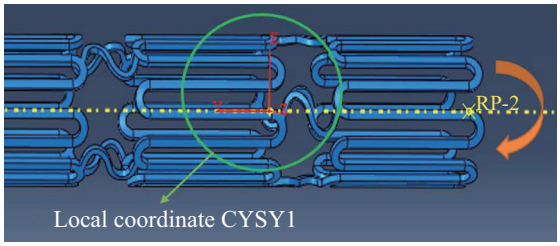


Fig. 5 the model for Cross-a: a local coordinate CYSY1 was build, and the angular displacement was applied on the stent around Z axis.

In order to evaluate the effect of bridge number on stent flexibility, the compressing, expanding and bending simulation was carried out on the 2 bridge stent, the 4 brigde stent and the 8 bridge stent. The steps, restraint,

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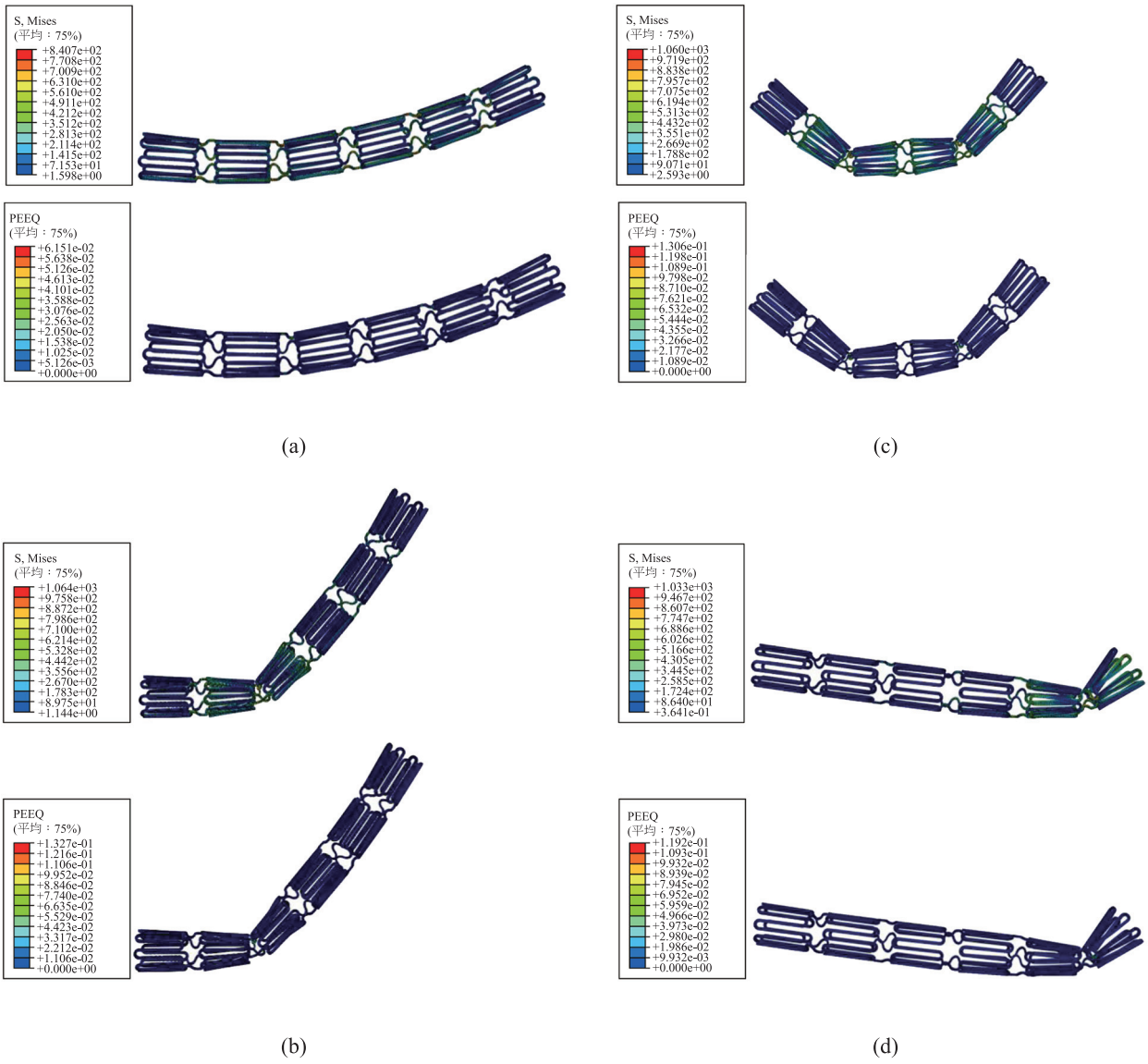


Fig. 6 Mises stress nephogram of bending deformation for 4 models: (a) Model 1, (b) Model 2, (c) Model 3 and (d) model 4.

and load used in Set C were the same as Set B.

2.2.4 Meshing

Since the full integral unit is prone to self-locking in bending deformation, a reduced integral unit (C3D8R) was adopted, and the unit is hexahedral. 3 layers of unites were arranged in the directions of bridge width and wall thickness. A value of 0.03 was specified as the element size for the global seeds, a value of 0.1 was specified as the default minimum size (as a fraction of the global element size). The final number of elements for the 2 bridge stent, 4 bridge stent, and 8 bridge stent were 39672, 57141 and 86328, respectively.

2.2.5 Interaction

A general contact interaction has been created, tangential behavior has been defined by penalty formulation, and the friction coefficient was 0.01.

Abaqus standard has been used to carry out all the simulations, and large deformation analysis was performed.

3. RESULTS AND ANALYSIS

3.1 Comparison of Four Mechanical Models for Bending Deformation of the Stent

3.1.1 Stimulation Results

Due to different loading methods, four different results of stent deformation were obtained, and their Von Mises stress nephograms were shown in Fig. 6.

Model 1 had only slight plastic deformation, the maximum Mises stress was 840 MPa and plastic deformation was 0.06. Models 2-4, due to different loading methods, the positions of large plastic deformation on the stent are different.

The bending stress of the stents was mainly concentrated on the plastically deformed bridges; In addition, the area of supporting rings connected to the bridges had high stress due to torsional deformation, as shown in Fig. 7. Torsional deformation in some supporting rings made the

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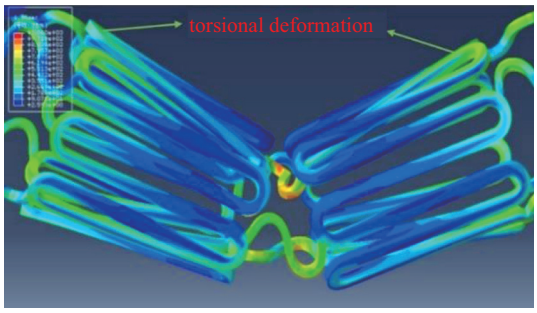


Fig. 7 Partial enlarged detail of Model 3: torsional deformation occurred on the rings.

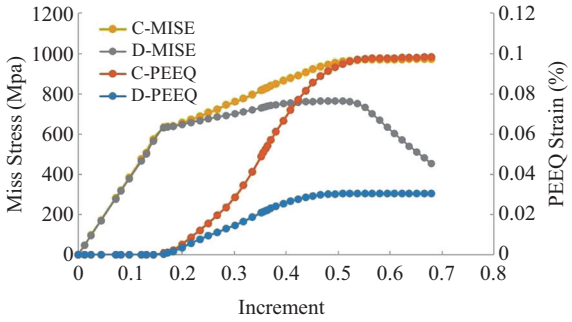


Fig. 9 Function of Miss stress and increment, function of PEEQ strain and increment of integral points C, D.

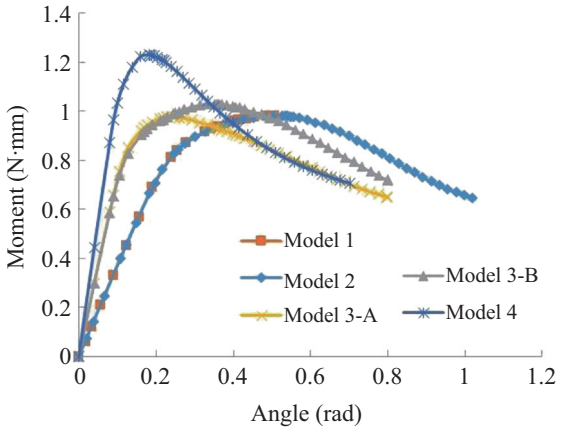


Fig. 8 Functions of moment and angle relative to Models 1-4.

stent loose cylindrical contours, while reducing the cross-sectional size, which would affect the adherence property of the stent and the blood flow characteristics at the lesion.

3.1.2 Flexibility Analysis

The flexibility of the stent is the reciprocal of the rigidity, and the slope of the moment-angle curve reflects the magnitude of the rigidity. The functions of moment and angle curves of the four models are shown in Fig. 8.

In Models 2-4, it was displacement load that applied on the control point, and the moment couldn't be directly read, so the reaction moment of the control point was outputted. In Model 3, the angular load was applied on both ends of the stent, and the function about reaction moment and angle at A and B were recorded. The two curves were consistent.

Models 2-4: Reaction moment increased linearly with the increase of angular displacement in the front steps. At this time, the stent was elastically deformed and the rigidity of the stent was constant. When the stent was deformed to a certain extent, some of the bridges were plastically deformed, the moment and the angle had a non-linear relationship, the stent rigidity decreased and the flexibility became better; due to the plastic deformation, reaction moment also decreased with the increase of the bending angle.

From the function curve of Model 1, the plastic deformation can't be observed. It is noticeable that the curve of Model 1 coincides with the front part of the Model 2 curve, which means that when the restraint is the same the calculated flexibility is equal for the stent with displacement load and the stent with moment load. Applying the displacement load, the whole process of bending deformation of the stent can be obtained.

The curves of Model 2, Model 3 and Model 4 don't coincide, but it doesn't mean that the rigidity of three models is different. For the given stent structure, the rigidity of the four models are calculated by Eqs. (2-5), respectively, and the results are 3.63 L, which are exactly the same.

Two different integral points were selected in the stent of the Model 2, and the function about stress and strain of the two points were shown in Fig. 9. Point C is on the position with large plastic deformation and the other integral point D is located in the small deformation area. Point C: the stress increase as the increasing of the load, but the increase speed slowed down in the plastic deformation stage. Point D: the stress increased slightly at the beginning of the plastic deformation of the stent, and with the increase of stent bending angle, the stress declined sharply. The plastic deformation caused the decrease of internal stress of the stent, thereby reducing the rigidity, as a result, the flexibility of the stent at plastic deformation stage is better than that at the elastic deformation stage.

The results of the stent flexibility calculated by four finite element models are consistent, but different models get different results of stent deformation. Compared with the moment-loading method, the displacement- loading method can reduce the calculation cost and help to understand the mechanism of bending deformation of the stent better. At the stage of elastic deformation, the flexibility index is constant; appropriate plastic deformation makes the stent flexibility higher.

Model 2 was used by Shobayashi for bending intracranial stents, and mechanical experiments were also conducted to verify the numerical simulation model. The results showed that the mechanical behavior of the stents was similar to that of the numerical simulations [22].

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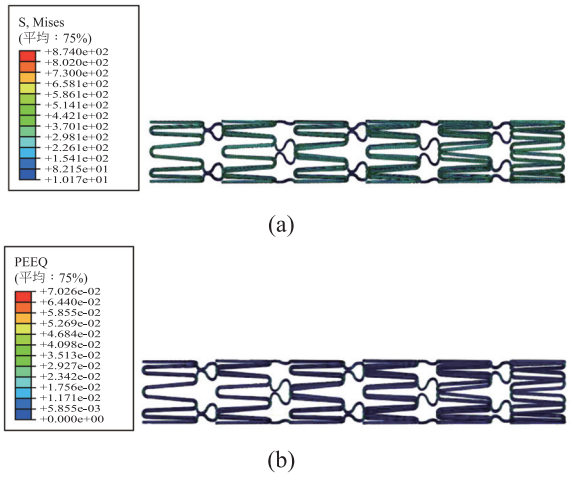


Fig. 10 Stress, strain nephograms of the stent with compressing, recoiling, expanding and recoiling deformation: (a) stress nephogram, and (b) strain nephogram.

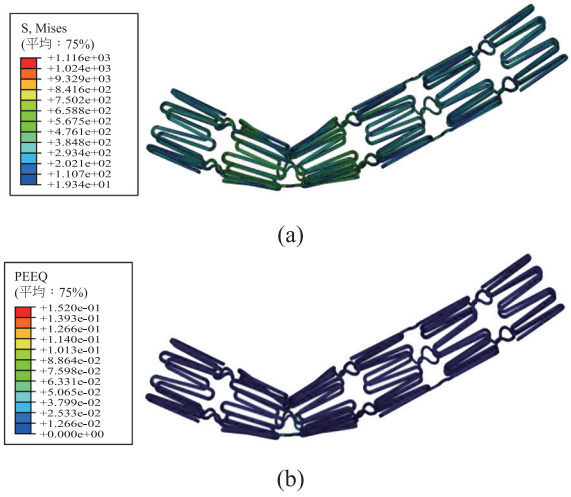


Fig. 11 Stress, strain nephograms of the stent with compressing, recoiling, expanding, recoiling and bending deformation: (a) stress nephogram, and (b) strain nephogram.

3.2 The Effect of Compressing and Expanding Deformation on Bending Rigidity

Stress and strain nephograms of the stent with compressing, recoiling, expanding and recoiling deformation are shown in Fig. 10; the maximum Mises stress is 887 MPa and the max PEEQ strain is 0.1009. The locations of the plastic deformation are mainly at the rounded angles of the supporting rings, and there are also a small amount of plastic deformation discovered on the bridges.

As shown in Fig. 11, with bending deformation, the maximum Mises stress increased to 1130 MPa and the max PEEQ strain reached to 0.156.

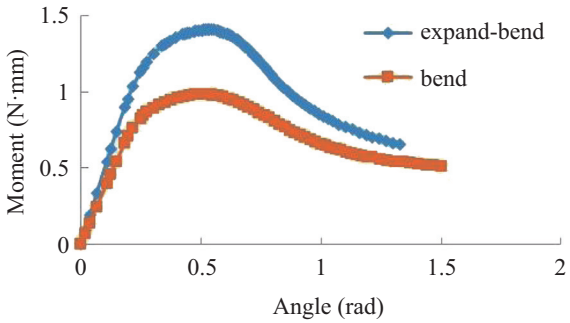


Fig. 12 Functions of moment and angle relative to expanded and non-expanded stents.

The function of moment and angle of the aforementioned stent is shown in Fig. 12, the overall shape is similar to the stent which was bended without expanding deformation. According to Eq. (3), the bending rigidity of the stent with expansion is 4.94 L, which is 1.36 times of the original stent, which means that the flexibility is 73.5% of that of the original stent. At the same time, according to Fig. 12, it can be found that there is a higher reaction moment inside the stent with expansion deformation. It means that the expanded stent have a stronger straightening effect on vessels.

The expansion deformation increases the diameter of the stent and increases the internal stress of the stent, which leads to the decrease of the stent flexibility. Taking the stent used in this paper as an example, the flexibility is reduced by about 26.5%; the maximum reaction moment is increased by about 42%. Therefore, when a flexibility test is carried out on the stent by using the finite element tool, more accurate results could be obtained by taking the effect of compressing and expanding deformation into consideration.

3.3 The Effect of Bridges on Flexibility

The bridges are the main factor that affects the flexibility of the stent. It has reached a lot of agreements with the effects of the shape and geometric parameters of the bridges on stent flexibility. But it is rarely reported that the effect of the arrangement of bridges on the flexibility. This paper will carry out studies with the focus on the arrangement of the bridges along the axial direction as well as the effect of the bridge number on the flexibility.

3.3.1 Cross and Non-Cross Arrangement of Bridges

The moment and angle relation curves of Cross-4 bridge stent and Noncross-4 bridge stent in different bending directions are shown in Fig. 13 and Fig. 14, respectively. The curve of Noncross-d is exactly the same as that of Noncross-a, thus not shown in the figure.

In the elastic deformation stage, the rigidity of the non-cross model is 12.83 L and the rigidity of the cross model is 4.94 L. The cross model shows excellent flexibility. But for a given stent, it shows no difference in flexibility at various bending directions.

In the plastic deformation stage, the flexibility of the

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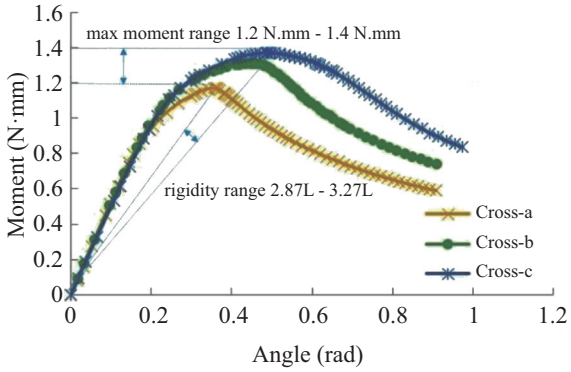


Fig. 13 Functions of moment and angle of Cross-4 bridge in different bending directions (the slope of the line connected by the highest point and origin represents the rigidity of stent in the plastic deformation stage).

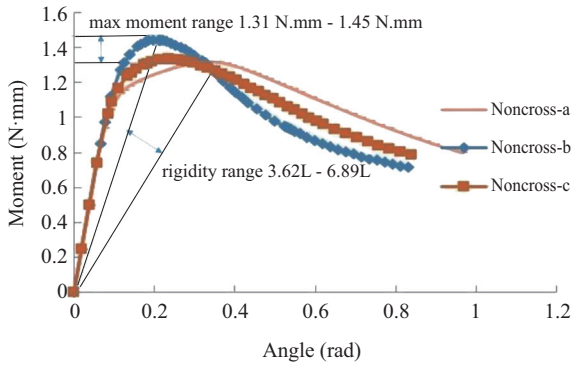


Fig. 14 Functions of moment and angle of Noncross-4 bridge in different bending directions.

stent shows difference in various directions. The range of maximum reaction moment of the stent with cross bridges is 1.2 N·mm-1.4 N·mm, the difference is not significant; but the bending angle corresponding to the maximum moment shows a large difference, if the rigidity of the stent is expressed as the slope of the line connected by the highest point and origin of the graph, then the rigidity range is 2.87 L-3.27 L. The range of maximum reaction moment of the stent with non-cross bridges is 1.31 N·mm-1.45 N·mm , which is slightly higher than that of the stent with cross bridges; and the rigidity range is 3.62 L-6.89 L, which is much higher than that of the cross stent model.

Therefore, the cross bridge can't effectively reduce the maximum reaction moment during the bending but can effectively improve the flexibility of the stent and the flexibility uniformity in different bending directions. At the same time, it can be seen from Fig. 13 and Fig. 14 that the maximum reaction moment is the minimum when bending the stent along the directions corresponding to Cross-a and Noncross-a, which can give some suggestions to the actual operation of stents implantation process.

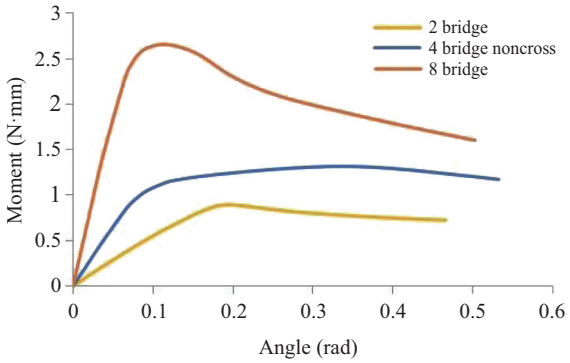


Fig. 15 Functions of moment and angle relative to the stents with 2 bridges, 4 bridges and 8 bridges.

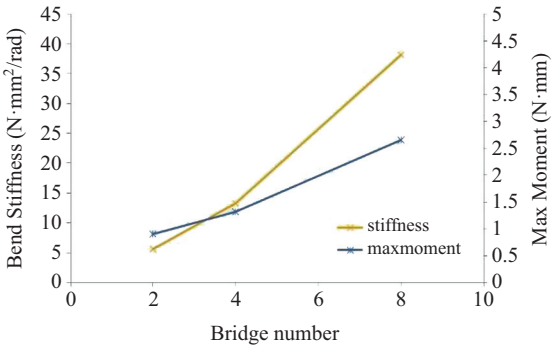


Fig. 16 Bend stiffness and bridge number index curve, max moment and bridge number index curve.

3.3.2 Bridge Number

The stents with 2 bridges, 4 bridges and 8 bridges were bent in the same direction, and the function of reaction moment and angle are shown in Fig. 15. In the elastic deformation stage, the bending rigidity of the three stents are 5.62 L, 12.83 L and 38.15 L, respectively, the bending rigidity increases nonlinearly with the increase of bridge number, as shown in Fig. 16. The maximum reaction moment also increases nonlinearly with the increase of bridge number. The bridge number has a significant effect on the flexibility of the stent, and its effect is much higher than that of bridge arrangement.

4. CONCLUSIONS

The bending simulation was carried out on the stents by the moment load and the rotational angle displacement load, as a result, the rigidity of the stents calculated were consistent. As for the moment-loading method, only the elastic process of bending deformation of the stent could be simulated; while the displacement-loading can stimulate the whole process of elastic as well as plastic deformation, which helps fully understand the mechanism of bending deformation of the stents and reduce the calculation cost. The same calculation results would be

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obtained for bending the stent with the mechanical model of the cantilever and the mechanical model of the simply supported beam.

The compressing and expanding deformation of the stent would increase the rigidity, therefore, when analyzing in the flexibility of stent, taking compressing and expanding deformation into consideration can get more accurate results. Different from the MPC multipoint constraint method, the method as coupling supporting rings at the both ends of the stent with the control points, could easily simulate the compressing-expanding-bending deformation of the stent.

The cross arrangement of bridges can effectively improve the flexibility of the stents and reduce the flexibility difference in various bending directions. The bridge number has a great effect on the flexibility of the stent, the stent rigidity and the max reaction moment in the bending process increase nonlinearly with the increase of bridge number.

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