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Performance Evaluation of the Vascular Model Based on the Nonlinear Viscoelastic Tensor-Mass Method

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***Abstract* - The virtual training system of vascular interventional surgery can improve the operation level of novice interventional surgeons and reduce the training cost. The deformation simulation of vascular tissue is the key part of the virtual training system for vascular interventional surgery. This paper presents a vascular tissue model based on nonlinear viscoelastic tensor-mass method (NVTMM), which can realize real-time visual feedback and haptic feedback. The hyperelastic model is used to simulate the nonlinear viscoelastic behavior of blood vessels. Meanwhile, the Euler's implicit integration scheme is used to make the simulation results more stable. In addition, a haptic force feedback device is introduced for the interaction between the vascular model and the virtual medical instrument, which improves the authenticity of the simulation. Finally, the real-time performance and accuracy of NVTMM, MSM and FEM are compared through experiments. The results show that NVTMM method has high real-time performance and can reflect the nonlinear viscoelastic behavior of blood vessels to a certain extent.**

Index Terms – Real-time simulation, HEML method, Vascular model.

I. Introduction

In recent years, with the acceleration of the pace of modern life, endovascular interventional surgery is more and more widely used in cardiovascular and cerebrovascular diseases. Minimally invasive vascular interventional surgery refers to the direct operation of the surgical guide wire and catheter under the guidance of medical images, which are sent to the lesion location (such as coronary artery, brain, liver and kidney). Then use the catheter to transport the drug or surgical instruments (such as balloon, stent, spring coil, etc.) to diagnose and treat the disease in vivo [1]-[3].

Endovascular interventional surgery has become a research hotspot in recent years because of its small trauma, strong targeting, small dosage and fast postoperative recovery. At present, many researchers at home and abroad are committed to the development of endovascular interventional surgery robot system to assist doctors in surgery. A vascular interventional surgical robotic system based on force-visual feedback was purposed in 2019 [4]. And in 2020, a pneumatic surgical robot for catheter ablation was designed [5]. In the same year, the research team of Beijing institute of technology proposed a non-contact detection method and designed a new type of main controller, which realized the real-time detection of the operation process without interference to the operation personnel [6]. A new tactile sensing robot assistant system based on magnetorheological fluid for vascular interventional surgery was developed by the research team of Kagawa University in 2021 [7]. Vascular interventional surgery robot generally adopts master-slave design, so that doctors are free from X-ray radiation in the operation process. However, the safe operation of the catheter and the positioning of the catheter tip on the target require doctors' technical skills and rich experience, which is difficult to obtain in a short time [8]-[12]. Therefore, there is a great demand for vascular interventional surgery training system.

The traditional training method of vascular interventional surgery is to train with the help of corpses or animals, but such resources are limited and expensive, which is difficult to meet the needs of training. At the same time, doctors need to wear heavy lead clothes and be exposed to X-ray for a long time during the traditional operation training, which poses a great threat to the health of doctors [13]-[17]. Considering these factors, the training system based on virtual reality is a promising way to help novice doctors carry out repeatable exercises in a risk-free environment. In addition, studies have shown that the use of virtual training system of vascular interventional surgery can improve the operation level of novice doctors [18], [19].

In the virtual training system of vascular interventional surgery, real-time and accurate simulation of vascular tissue deformation is a challenging problem. At present, the most common way to simulate vascular tissue deformation is physical based soft tissue deformation method, which can be divided into two kinds: mesh based and meshless modeling methods.

Mesh based modeling methods include mass spring method, finite element method and mass tensor method [20]-[22]. Mass spring method (MSM) has the advantages of fast simulation speed and high real-time performance, but its accuracy is low, system stability is poor, and it is unable to simulate large deformation. Finite element method (FEM) has very high simulation accuracy and can provide more real deformation. However, the computation of finite element method is too large to meet the real-time requirements of virtual surgery. Mass tensor method is a combination of finite element method and mass spring method, which is easy to achieve large deformation, but it will consume a lot of computer memory when it involves complex tetrahedral mesh, and the computational efficiency is low under nonlinear conditions. The commonly used meshless modeling methods include weighted residual collocation method (WRPCM), particle hydrodynamics (SPH), moving least squares (MLS), radial basis function (RBF), element Galerkin (EFG), etc. Among them, EFG method has better accuracy, convergence and stability, and is the most complete meshless method [23]. Meshless method can predict soft tissue deformation without involving the mesh topology of discrete tissue model. It is often used to simulate the cutting, tearing and other behaviors of soft tissue. However, this method requires the calculation of node to node adjacency in each time step, which leads to additional computational load.

This paper focuses on the real-time numerical simulation of blood vessel model and large deformation problem. In our master-slave virtual training system of vascular interventional surgery, nonlinear viscoelastic tensor-mass method (NVTMM) is used to simulate vascular tissue. This method is not only convenient for fast calculation, but also can achieve the refresh rate of 100 Hz visual feedback. Due to the vascular tissue shows nonlinear viscoelastic response in the experiment [24]. In order to simulate the mechanical response of biological soft tissue more truly, a viscoelastic hyperelastic constitutive model based on Prony series is selected. In addition, the mass spring model and the finite element model are implemented on the tetrahedral mesh to compare and verify the advantages of the NVTMM in the real-time and authenticity of the simulation. Finally, the vascular model and guide wire model are coupled to form a master-slave virtual training system with force feedback and real-time interaction.

The rest of this paper is arranged as follows: The second part gives an overview of the NVTMM method. In the third part, the scheme of introducing viscoelastic properties into NVTMM is introduced. In the fourth part, the corresponding experiments are designed to verify the real-time and authenticity of the NVTMM, and compared with MSM and FEM. In the fifth part, the experimental results are discussed and some conclusions and future work are put forward.

II. Nonlinear Viscoelastic Tensor-Mass Method

The blood vessel model should be a physical real soft tissue mechanical model. Therefore, the model should be as realistic as possible and effective enough to allow real-time performance. Compared with spring mass method and finite element method, NVTMM is more suitable for this requirement. Due to the linear elastic model is only suitable for small displacement, the Neo Hookean hyperelastic constructive model is used to simulate the nonlinear deformation of vascular tissue.

In the NVTMM, the simulated object is discretized into a P1 tetrahedral mesh defined by the finite element theory, as shown in Figure 1. The P1 element has the property of equal deformation gradient tensor on a given tetrahedron, which is very efficient in calculating the force field. As the vascular tissue and other biological soft tissue materials are generally

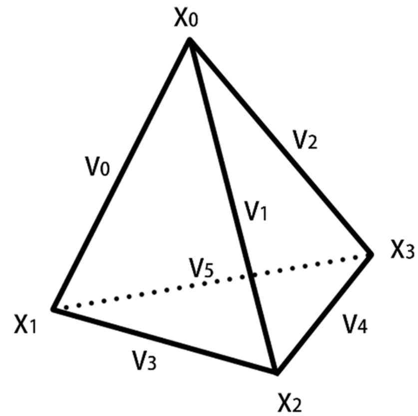


Fig. 1 P1 tetrahedral mesh

considered to be super elastic. This kind of material may have large deformation. In order to describe the geometric transformation of this kind of object, the deformation gradient tensor F is introduced：

$$F = I + \nabla U \quad (1)$$

where I is the unit tensor and U is the displacement vector.

The strain tensor ε depends on the deformation gradient defined by Formula (1) and can quantify the deformation of the object. Moreover, the deformation of the object under the action of external force will produce an elastic potential energy, which depends on the constitutive equation of the material used. And the corresponding deformation force can be obtained by the derivative of the energy. Therefore, the motion of the deformable body can be expressed by the following differential equations:

$$M\frac{d^2U}{dt^2} + D\frac{dU}{dt} + KU = F \quad (2)$$

where M is the mass, D is the damping and K is the stiffness matrix.

According to the expected mechanical behavior of materials, there are different elastic models. These models are defined by the strain tensor formula ε. Because the deformation of vascular tissue is nonlinear, the nonlinear elastic behavior must be considered. Hyperelasticity provides a way to simulate this material. The simplest hyperelastic model is an extension of the linear elastic material based on the Green-St Venant strain-tensor ε_{nl}:

$$\varepsilon_{nl}(X) = \frac{1}{2}(\nabla U^T(X) + \nabla U(X) + \nabla U^T(X)\nabla U(X)) \quad (3)$$

The energy of deformation W_{nl} is then defined by:

$$W_{nl}(X) = \frac{\lambda}{2}(tr\varepsilon_{nl}(X))^2 + \mu tr(\varepsilon_{nl}(X)^2) \quad (4)$$

where λ and μ the Lame coefficients characterizing the material stiffness, defined by:

$$\lambda = \frac{\upsilon E}{(1+\upsilon)(1-2\upsilon)},\ \mu = \frac{E}{2(1+\upsilon)} \quad (5)$$

where E is the Young's modulus and υ is the Poisson's ratio.

And the deformation energy of any tetrahedral element in the discrete domain can be given by the following formula:

$$W_E = W_{nl}(X)\frac{1}{Vol_0} \quad (6)$$

where W_E is the deformation energy of tetrahedral element, Vol_0 is the volume of the tetrahedral element.

Then the force applied on the node P_i can be given by:

$$F(P_i) = \sum_{k\,|\,E_k\in E_{p_i}} F_{E_k}(P_i) = \sum_{k\,|\,E_k\in E_{p_i}} \frac{\partial W_{E_k}(P_i)}{\partial U_i} \quad (7)$$

And in order to ensure the stability of the simulation, Euler implicit integration scheme is used. The Euler implicit integration scheme needs derivative force, so the derivative force also needs to be calculated:

$$dF^j(P_i) = \sum_{k\,|\,E_k\in E_{P_i}} \frac{\partial F_{E_k}(P_i)}{\partial U_j} \quad (8)$$

where the E_{P_i} is the set of elements containing node Pi.

Neo Hookean is one of the most common hyperelastic material models, which can be used to predict the nonlinear stress-strain behavior of rubber or biomechanical materials under large deformation. Neo Hookean model is derived from the classical results of statistical thermodynamics. Compared with other hyperelastic models, Neo Hookean model is simple, efficient and versatile. For Neo Hookean model, its energy density $\Psi(F)$ can be expressed as:

$$\Psi(F) = \frac{\mu}{2}(tr(F^T F) - d) - \mu\log(J) + \frac{\lambda}{2}(\log(J))^2 \quad (9)$$

where F is the deformation gradient tensor, D is the dimension, and J is the reciprocal of the elastic modulus, that is, the deformation rate per unit stress of the material.

III. Visco-hyperelasticity Constitutive Model

Viscoelasticity is an important property of biological tissue, which shows mechanical behavior changing with time. The viscous effect caused by viscoelasticity can not be ignored for accurately and truly describing the mechanical properties of tissue. In order to fully describe the mechanical response of biological soft tissue, a viscoelastic hyperelastic constitutive model based on Prony series is established, and a time-dependent term is added to the basic hyperelastic formula. The time-dependent strain energy function can be expressed in the form of convolution time integral:

$$\widehat{W}(W_E, t) = \int_0^t \alpha(t-s)\frac{\partial W_E}{\partial s}ds \quad (10)$$

where t is the time. And the relaxation functions α(t) is given by a Prony series:

$$\alpha(t) = 1 - \sum_{i=1}^{n} \alpha_i(1 - e^{-t/\tau_i}) \quad (11)$$

where n is the number of Maxwell elements, α_i is the ith normalized modulus of elasticity and τ_i is the relaxation time.

Hence, the visco-hyperelastic force $\hat{F}$ can be written as:

$$\hat{F} = -\frac{\partial \widehat{W}}{\partial x} = -\int_0^t [1 - \sum_{i=1}^{n}\alpha_i(1 - e^{(s-t)/\tau_i})]\frac{\partial F}{\partial s}ds \quad (12)$$

Equation (12) can also be written as:

$$\hat{F} = -(F - \sum_{i=1}^{n}\gamma_i) \quad (13)$$

$$\gamma_i = \int_0^t \alpha_i(1 - e^{(s-t)/\tau_i})\frac{\partial F}{\partial s}ds \quad (14)$$

After discretization of time, the integral equation can be converted into incremental updating formula:

$$\gamma_i^n = A_i F^n + B_i \gamma_i^{n-1} \quad (15)$$

where $A_i = \Delta t\alpha_i/(\Delta t + \tau_i)$ and $B_i = \tau_i/(\Delta t + \tau_i)$. Δt is the time step and the superscripts are added to indicate time increment.

IV. Experiments and Results

The vascular interventional surgery training system used in this paper is built on a personal laptop, equipped with 16 GB memory, Intel i7-10750h processor and NVIDIA rtx2060 graphics card. The verification simulation is carried out on sofa medical simulation engine. The sofa simulation engine used is version 20.06, and the required supporting software versions are visual studio 2017 and python 2.7. The vascular tissue model is a tetrahedral mesh model, which is obtained by 3D modeling after simplifying the aortic arch of the patient. The tetrahedral mesh model of the aortic arch is shown in Figure 2. The guide wire is a cylinder model generated by Maya modeling software.

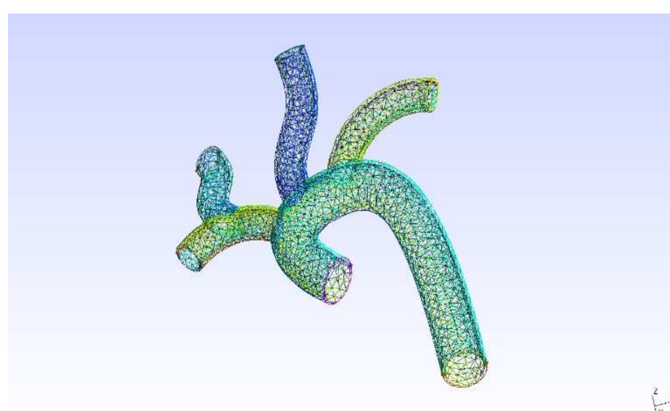


Fig.2 The tetrahedral mesh model of aortic arch

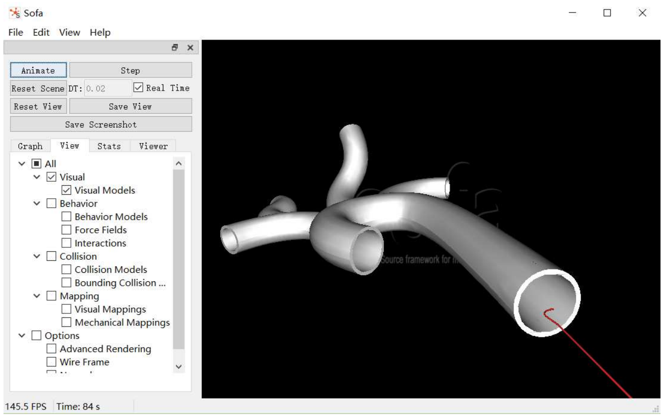


Fig.3 Sofa engine simulation of vascular interventional surgery interface

The experimental platform is mainly divided into two parts: simulation environment based on virtual reality and force feedback equipment, as shown in Figure 4. In this figure, A is the virtual reality simulation environment based on sofa framework, B is the force feedback equipment. The force feedback device used is Geomagic touchX, which can provide precise positioning input and high fidelity force feedback output. The pose information obtained by the force feedback device will be transmitted to the virtual reality simulation environment to control the movement of the virtual guide wire. In the virtual reality simulation environment, the interaction between the virtual guide wire and the vascular tissue will be simulated, and the feedback force obtained from the simulation will be output to the force feedback device through the tactile rendering method.

A. Experiments of the Real-time Performance

Real time is an important performance index of vascular interventional surgery simulator based on virtual reality. The level of real-time directly affects the interaction and immersion of the simulator. In the experiment, the visual refresh rate and tactile refresh rate of the whole virtual surgery training system are tested when NVTMM method is used to simulate vascular tissue with different number of nodes. In the experiment, the force interaction between the guide wire and the vascular wall was simulated by controlling the guide wire or mouse to exert

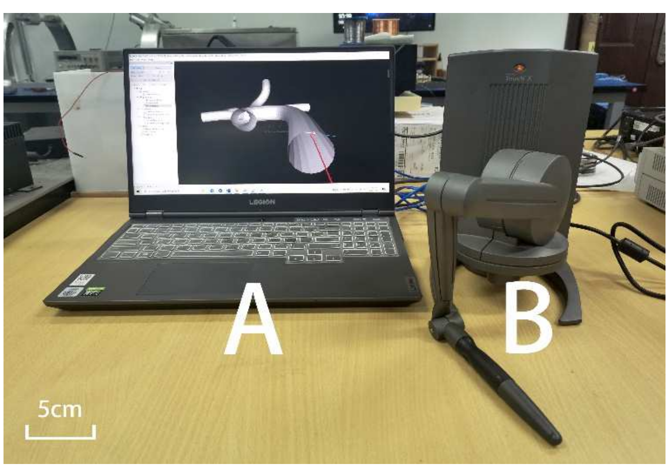


Fig. 4 overview of the experimental platform

TABLE I
THE VISUAL REFRESH RATE OF DIFFERENT MODELS WITH DIFFERENT NUMBER OF NODES

Number of nodes	MSM(FPS)	FEM(FPS)	NVTMM(FPS)
1458	276	94	103
5832	70	26	55
23328	18	4	13

TABLE II
THE TACTILE REFRESH RATE OF DIFFERENT MODELS WITH DIFFERENT NUMBER OF NODES

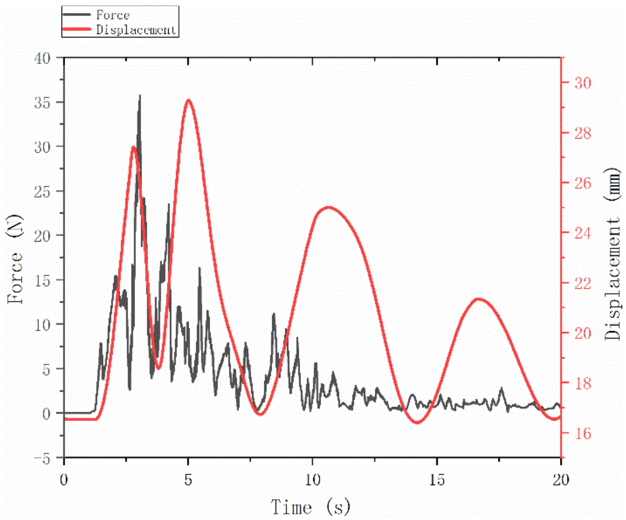
Number of nodes	MSM(FPS)	FEM(FPS)	NVTMM(FPS)
1458	142	87	124
5832	33	25	58
23328	6	4	12

external force on the vascular model. Then, the MSM and FEM were used to simulate the vascular tissue and test the real-time performance of the above methods. The visual refresh rate and tactile refresh rate of different models with different number of nodes are shown in Table 1 and table II.

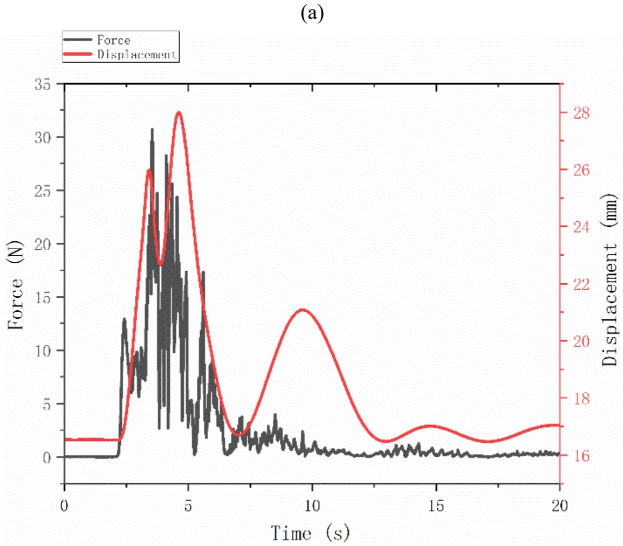
The experimental results show that the computational efficiency of NVTMM is higher than that of MSM and FEM, while the visual refresh rate of NVTMM is higher than that of FEM and slightly lower than that of MSM. FEM obtains the global solution by solving the equations on each discrete element and combining the results into a global matrix. In this method, the physical characteristics are directly integrated into a mechanical formula, so the deformation can be simulated with high accuracy. But this method also leads to the need for more computation time when dealing with nonlinear behavior. NVTMM computes locally and iteratively for each discrete element, so it has higher computational efficiency than FEM. In addition, this allows for more direct and easier handling of topology changes and external interactions. Although the visual refresh rate of MSM is higher than that of NVTMM, the parameters in MSM model have no real attributes, which can not be combined with the real biomaterial parameters, and can only be simulated in specific cases. Moreover, MSM can not simulate the nonlinear deformation of objects. Therefore, considering the generality and real-time of the algorithm, NVTMM is more suitable for operation simulator.

B. Experiments of the Viscoelasticity Performance

In the vascular interventional surgery simulator based on virtual reality, the vascular tissue model should not only have high real-time performance, but also simulate the biomechanical characteristics of vascular tissue as real as possible. In order to verify the simulation of viscoelasticity by NVTMM, a control experiment is designed. In this experiment, the NVTMM model and the TMM model were applied with the same force. Fig. 5 (a) and (b) record the absolute values of force and displacement of the two models at the 278th node, respectively. The red line in Figure 5 records the relationship between displacement and time, while the black line records the relationship between force and time.



(a)



(b)

Fig.5 The force and displacement response of the 278th node under the constant force. (a) the displacement and force response of TMM model. (b) the displacement and force response of NVTMM model.

It can be seen from the experimental results that a constant force is applied to the model when t = 3s. TMM model does not have viscoelasticity, so it needs a long time to be stable because of its large amplitude of vibration. Because of the Prony series, NVTMM model will produce lag phenomenon after being forced. The specific performance of NVTMM model is that the amplitude of vibration is smaller after being forced, and it can tend to be stable at a faster speed.

V. Conclusions

In this paper, NVTMM is used to simulate the vascular tissue, which is derived from the finite element method and can simulate a variety of super elastic materials with bio viscoelasticity. Experiments show that, compared with the classical FEM and MSM, NVTMM has the advantages of high computational efficiency, and can realize real-time visual feedback and tactile feedback. Under the condition of ensuring 30Hz visual refresh rate and 100Hz tactile refresh rate, the NVTMM can deal with the model with higher number of faces, so as to improve the simulation accuracy to a certain extent. It can be seen from the above experiments that the NVTMM takes into account both the accuracy of biomechanics and the computational efficiency and can effectively improve the overall performance of the virtual interventional surgery simulation system. In the next step, we will expand the NVTMM to simulate heterogeneous and anisotropic materials, so as to better simulate the biological characteristics of vascular tissue.

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