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Evaluation and User Study of Haptic Simulator for Learning Palpation in Cardiovascular Surgery

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Abstract

This study presents a haptic simulator for learning palpation of aorta in cardiovascular surgery and performs quantitative evaluation in educational use through some user study. The simulator implements physics-based methods that enable VR simulation of soft organs with autonomous motion like a beating heart or an aorta. The developed model simulates real-time deformation and force feedback during surgical palpation based on finite element methods. The evaluation with cardiovascular surgeons and medical students confirms the developed system is useful for learning not only surgical procedures but also stiffness of in-vivo organs.

Key words: Surgical Training, Virtual Human, FEM-Based Modeling, Palpation, Medical Education

1. Introduction

Virtual reality (VR) has been utilized in various fields of medicine and opens new possibilities in areas of surgical training and preoperative planning. Surgical simulators with haptic feedback have been developed at many research institute and companies to create environments that help train surgeons or students [1, 2].

Many surgeons specifically in cardiovascular surgery, however, still make great efforts in learning most surgical procedures and training of their skills. Although intraoperative palpation of aorta is a key procedure to grasp sclerosis region and status, no training environment is given and surgeons have to acquire the skill through conventional “learning by doing” approach.

In order to establish haptic simulators in cardiac surgery, modeling of virtual organs like human heart or aorta is essential. The model requires not only considering autonomous beating of organs but also presenting visual and haptic feedback in response to surgeons’ manipulation. Since these requirements are technically severe in accurate modeling and fast computation, practical systems are not currently produced.

Also, current VR simulators [1, 3] mostly handle simple objects that have uniform elastic parameters and try to represent approximate tissue behavior. Because the palpation simulator has to enable surgeons to learn difference of stiffness caused by beating and sclerotic diseases, more detailed soft tissue modeling based on elastic theory is required. Further examination observing effects of simulators and stiffness recognition is also needed for practical use of haptic simulators in medical education or in preoperative rehearsal.

This study proposes a new haptic simulator for residents or medical students to learn palpation of active organs that have autonomous motion. The simulator implements a beating aorta model, which simulates real-time deformation and force feedback during surgical palpation based on finite element method (FEM). We provide a C++ library of the model and prepare pre-processing software for flexible finite element modeling.

Unlike most forgoing study, this paper gives quantitative evaluation of the developed simulator in educational use. Effectiveness of practicing procedures through haptic feedback using virtual organs is confirmed through several experiments and user study with skilled surgeons and medical students.

2. Haptic Simulator for Palpation of Aorta

This chapter gives a design of the proposed haptic simulator for learning palpation of aorta. The authors focus on intraoperative palpation to grasp sclerosis status and extract the following functions that should be provided by the simulator.

- Real-time calculation of accurate deformation and reaction force
- A beating aorta model that reflects pressure and partially different stiffness

- Enabling users to pinch the virtual aorta with at least two fingers

Based on these requirements, this study designed the haptic simulator as shown in Fig. 1. The position of a manipulated point (manipulator) is given through haptic devices. The manipulator represents a tip of fingers or surgical instruments in the virtual environment. If collision detection scheme detects interaction between the manipulator and the virtual heart object, both the contacted point and its displacement are applied to the physics-based aorta model, which provides visual and haptic feedback to the user.

To satisfy the first requirement, this paper gives a finite element based computational model of aortas. The model calculates reaction force and biomechanical deformation reflecting internal pressure in response to surgeon's manipulation. The second requirement is fulfilled by a developed modeling tool with efficient pre-processing techniques. Measured blood pressure is applied to the developed model. Lastly, implementation and registration of two haptic devices enables pinch manipulation. The details of our approach are described in chapter 3, and results and quantitative evaluation of the developed simulator are given in chapter 4.

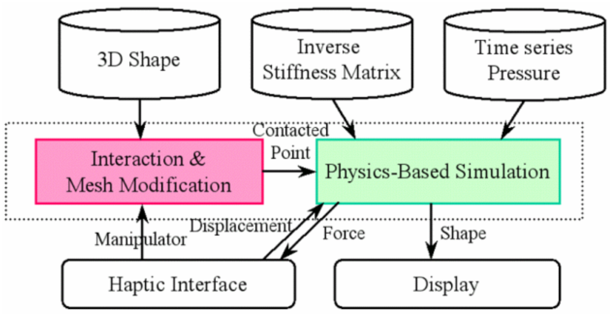

Fig. 1: Overall simulation framework of the developed haptic simulator for learning palpation of aorta.

3. Physics-Based Modeling of Beating Aorta

3-1. Construction of Arterial Sclerosis Model

Physics-based simulation requires a 3D shape with elastic information of the target organ. CT or MRI images acquired form patients are used to construct a virtual aorta model. 3D region of the aorta is extracted from the voxels and divided into finite tetrahedra. Each tetrahedron represents a part of the 3D region of the aorta and also has physical parameters: young modulus and poisson ratio. The volumetric mesh topology and stiffness parameter are useful to yield a stiffness matrix for finite element based simulation.

This study represents arterial sclerosis by partially setting high young modulus to the sclerosis region like Fig.2.

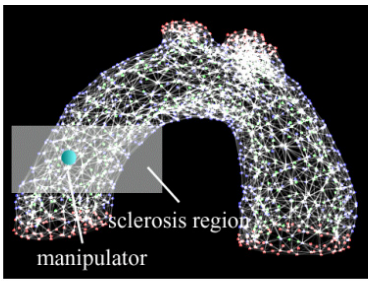

Fig. 2: An aorta sclerosis model. Tetrahedra in sclerosis region have high young modulus.

The 3D sclerosis region is acquired from CT or MRI images. Since stiffness parameter of in-vivo aortas cannot be measured non-invasively, this study prepares a modeling tool: *MatrixBuilder*, which enables flexible parameter setting via its graphical user interface. The valid stiffness setting for normal and sclerosis region is examined later through characteristic evaluation with several expert surgeons.

The *MatrixBuilder* also enables all pre-processing procedures to construct the stiffness matrix for real-time finite element based simulation. The procedures are listed as follows.

(a) Physical condition setting
Some physical conditions (e.g. fixed condition and boundary condition) on specific vertices are defined.

(b) Stiffness matrix generation
The stiffness matrix K is calculated based on tetrahedral mesh topology, stiffness parameters and given physical conditions.

(c) Reduction of stiffness matrix
The size of the stiffness matrix is reduced by condensation technique [4, 5]. Next, matrix elements for fixed vertices are eliminated because they are not used to finite element based simulation.

(d) Calculation of inverse stiffness matrix
The inverse matrix L of the condensed stiffness matrix K is calculated. The constructed inverse stiffness matrix L is useful to real time simulation.

5-2. FEM-Based Computation

Both a haptic rendering scheme and physics-based visual and force feedback for the constructed aorta model perform virtual palpation. A manipulator is modeled as a point and collision is detected using the point based haptic rendering method [6, 7], which enables users to push surfaces of the virtual object. After collision detection, deformation and reaction force are calculated in real time using a finite element based computation model that handles dynamic blood pressure of aortas. Fig. 3 briefly illustrates process flow of the proposed model using 2D outline of the aortic wall.

When blood pressure is applied to the aortic wall, finite element based simulation provides an expanded shape (Fig. 3-a). If the higher pressure is applied, the expansion becomes larger (Fig. 3-b). These results show autonomous beating of aortas. In addition, if external force gives small displacement in the aorta model, reaction force is calculated and conveyed to user's hand via haptic interfaces (Fig. 3-c). Because surgeons do not pinch the wall strongly in actual surgery, the proposed model assumes that blood pressure is not affected by surgical palpation.

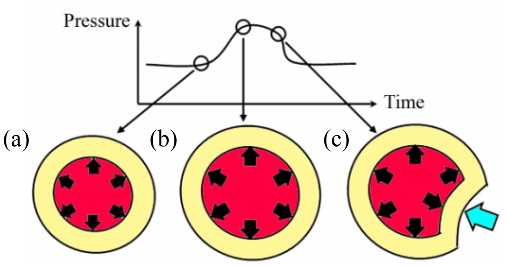

Fig. 3: Finite element based computation model for a beating aorta. 3D shape, stiffness and blood pressure are applied. Surgical palpation can be performed with autonomous beating.

Next, the authors explain calculation methods of the proposed deformable model based on linear elastic theory. Assuming that internal force is in equilibrium at each discrete time, the relationship between external force and displacement on an elastic object is defined by the following equation:

$$f = Ku \tag{5.1}$$

where on all vertices, u is displacement and f is external force including blood pressure. K denotes stiffness matrix of the object. Note that K matrix can be efficiently reduced by condensation [4, 5] in the preprocessing stage. Equation (5.1) is simplified to equation (5.2) using the condensed inverse stiffness matrix L, which defines physical relationship between external force f and displacement u on surface vertices.

$$u = Lf \tag{5.2}$$

The surface vertices are divided into three groups: contacted vertices, inside wall vertices and other free vertices. Contacted vertices are directly displaced by user's manipulation. Inside wall vertices are dynamically affected by blood pressure. Equation (5.3) is given as the detailed description of equation (5.2) using initial letters of categorized vertices.

$$\begin{pmatrix} u_i \\ u_o \\ u_c \end{pmatrix} = \begin{pmatrix} L_{ii} & L_{io} & L_{ic} \\ L_{oi} & L_{oo} & L_{oc} \\ L_{ci} & L_{co} & L_{cc} \end{pmatrix} \begin{pmatrix} f_i \\ f_o \\ f_c \end{pmatrix} \tag{5.3}$$

where f_i denotes blood pressure that is applied to inside wall of the aorta, and u_c is displacement of the contacted vertex. Considering that f_o is constant zero, f_c is given as the following equation:

$$f_c = L_{cc}^{-1}\left(u_c - L_{ci} f_i\right) \tag{5.4}$$

f_c is external force on the contacted vertex and denotes reaction force that is conveyed to the user. Applying f_c to the equation (5.3), displacement u_o of other free surface vertices can be acquired. Dynamic transition of u_o due to time series blood pressure shows autonomous motion of beating aortas.

The overall calculation methods are integrated into C++ library, which is provided through web page at Kyoto University Hospital..

4. Evaluation and Results

This chapter demonstrates several results performed by the developed palpation simulator, and gives some evaluations on performance and validity of the developed system. The overall algorithms were installed on a standard PC (CPU: Pentium III Dual 933MHz, Memory: 2048MB, Graphic Card: GeForce 3 and OS: Windows 2000). Two PHANToMs (Sensable Technologies) were implemented as force feedback interfaces. For fast calculation of the stiffness matrix, Intel Math Kernel Library [8] was implemented.

A 3D shape of a normal aortic arch was extracted from the Visible Human Male [9]. The number of the vertices of tetrahedral mesh is 1651. For verification of the system performance, stiffness parameters of the elements are fixed as uniform values. The young modulus is 1.0MPa and the poisson ratio is 0.48. The valid stiffness parameter is configured through characteristic evaluation with cardiovascular surgeons.

Fig. 4 demonstrates deformation results in pinching a virtual aorta by two fingers using the developed palpation environment. Two sphere images denote the current positions of the manipulators, which are operated through tips of two PHANToMs. The finite element based computation model calculates reaction force and deformation reflecting time series aorta pressure in real time.

Fig. 4: Real time deformation in palpating a virtual aorta. Two manipulators enable users to pinch the aorta interactively.

4-1. Validation of Force Feedback Model

The authors give an experiment with 8 cardiovascular surgeons in order to examine whether the system represents physical behavior of in-vivo aortas. 20 aorta models with the same 3D shape and different stiffness are prepared. From 0.2MPa to 4.0MPa is given to the models as uniform stiffness.

Surgeons palpate the 3D virtual aorta models and select the most realistic one that simulates stiffness of normal aortas. They also determine threshold stiffness that should be regarded as sclerosis status in actual surgery. Pinching manipulation is performed by attaching two fingers to the joints of two PHANToMs using rubber bands like in Fig. 5.

Fig. 5: Pinching manipulation via two PHANToM Desktops. Two fingers are attached at the joints of the PHANToMs using rubber bands.

Fig. 6 (a) presents statistical results of selected normal stiffness by 8 cardiovascular surgeons. The graph shows surgeons select only two patterns: 1.0MPa and 1.2MPa. This result suggests following two significant views:

- The developed model whose young modulus is nearly 1.0MPa efficiently simulates stiffness of in-vivo normal aortas
- Skillful surgeons recognize absolute stiffness of normal aortas

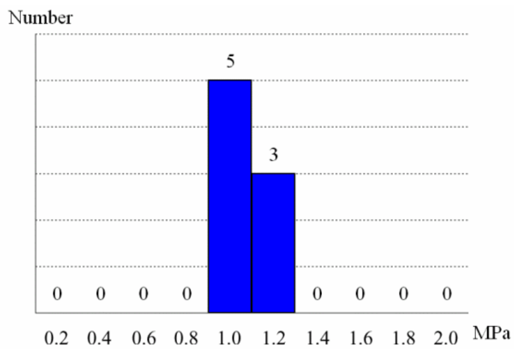

Fig. 6 (a): Characteristic configuration of normal aorta's stiffness by skilled surgeons.

Also, Fig. 6 (b) shows characteristic configuration of threshold stiffness of sclerosis using the virtual aorta model. The graph suggests surgeons estimate the region that has over 3.0MPa stiffness is sclerosis.

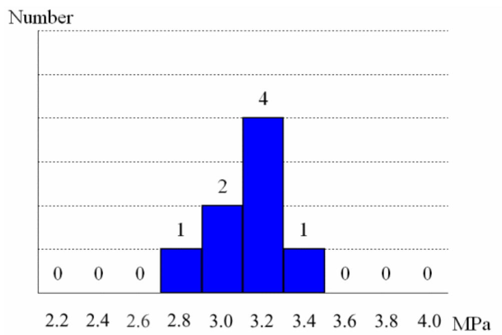

Fig. 6 (b): Characteristic configuration of threshold stiffness between normal and sclerosis case.

To confirm dynamic force feedback quantitatively, reaction force is measured under the condition that the manipulator is placed on the surface and fixed against expansion of the aorta wall. A sclerosis aorta model is prepared using specific sclerosis region shown in Fig. 2. 6.0 MPa is applied to the stiffness of sclerosis region. Fig. 7 shows relationship between dynamic transition of reaction force and time series blood pressure. Both absolute value of reaction force and beating status are differently represented at the two regions, which realistically simulate physical behavior of normal and abnormal wall of aortas.

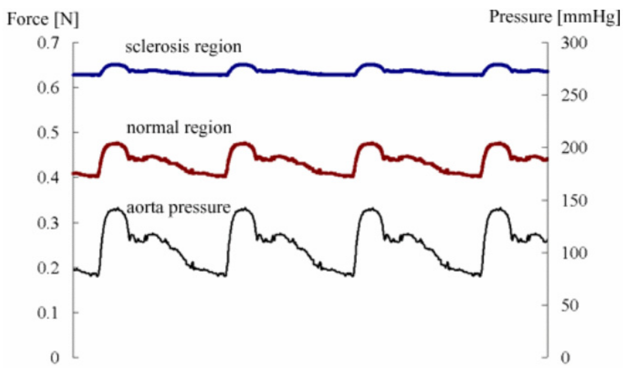

Fig. 7: Reaction force in pushing sclerosis and normal region of a virtual aorta sclerosis model.

4-2. Analysis of Stiffness Recognition

As shown in the previous experiments, stiffness recognition of normal and abnormal aortas is an essential skill for surgical palpation. To confirm the developed system is useful for training and rehearsal of surgical palpation, the authors examine stiffness recognition of virtual aorta models with 18 medical students who have not experienced palpation of in-vivo aortas.

The 10 virtual aorta models with 0.2MPa - 2.0MPa stiffness used in previous experiments are prepared. Medical students select one model under the following three conditions as to elastic information.

Condition 1:
No information. Medical students know only the stiffness of cadaver's aortas, and therefore have to use their imagination in order to select the stiffness of in-vivo aortas.

Condition 2:
Verbal information. Some cardiovascular surgeons express stiffness of normal aortas using stiffness of a rubber hose. This conventional teaching method is given.

Condition 3:
Force feedback. Medical students can learn the stiffness of normal aortas via the palpation system in a few minutes. The stiffness of the normal aorta model is 1.0 MPa.

Fig. 8 shows statistical results of the selected stiffness. The F-test result shows there is significant difference between condition 1 and condition 3. Also, difference is found between condition 2 and 3. As shown in Fig. 8, the two graphs of condition 1 and condition 2 are dispersed. This result confirms verbal information is not effective to learn the stiffness of the aorta model. The graph of condition 3 is nearly based on normal distribution and more than 80% students indicate from 0.8 MPa to 1.2 MPa models. This result demonstrates that the developed system is useful to recognize the normal stiffness of the virtual aorta model. Because the young modulus 1.0 MPa is configured in the previous experiments with expert surgeons, the students efficiently experience and obtain the stiffness information of in-vivo normal aortas through virtual interaction.

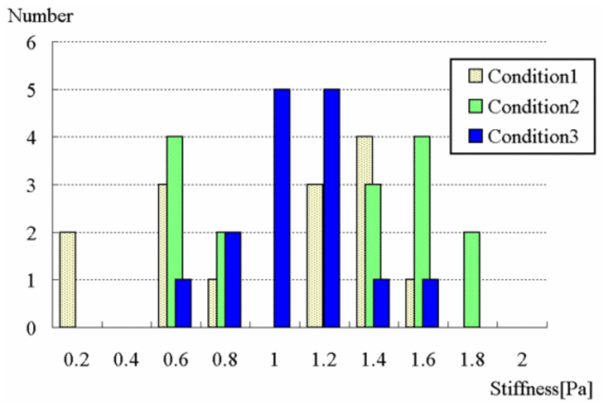

Fig. 8: Stiffness recognition of 1.0 MPa normal aorta model in different cases: conventional instruction (condition 1 or 2) and simulator based instruction (condition 3).

4-3. Verification of Identifying Sclerosis Region

The last experiments aim to confirm whether or not the developed system makes possible of representing local stiffness of a sclerosis region. The authors give three sclerosis models, and medical students search the sclerosis region through virtual palpation. The students indicate normal region where a surgical clip can be set near the boundary between normal and sclerosis region. The stiffness of normal region is 1.0MPa and sclerosis region is 5.0MPa.

Fig. 9 shows pointed positions on the three sclerosis aorta models. The colored area is the sclerosis region. Most of the students can indicate the normal region and can identify the difference between normal and sclerosis status. Thus, haptic feedback is useful to rehearse surgical palpation of these sclerosis cases. These figures also suggest that some students cannot recognize the boundary, which shows some significant aspects. Further discussion with expert cardiovascular surgeons is given in the next section.

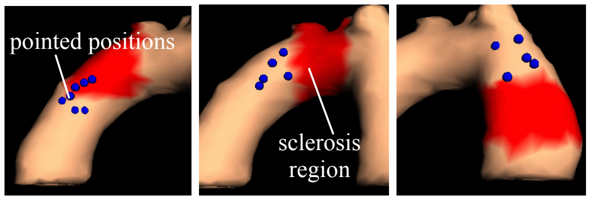

Fig. 9: Pointed positions on the two aorta sclerosis models by medical students. The difference between normal and sclerosis status is identified.

5. Discussion

Lastly, the authors discuss the proposed system and possibility of further improvement for providing realistic virtual reality based palpation. First of all, in order to achieve real time refresh rate, the authors simplified physical behavior of in-vivo aorta and extracted some essential functions. Either difference of local pressure or blood flow is not presented. Autonomous beating is simulated using sequential blood pressure. If 3D shapes of aorta models are not complex, the blood pressure applied to the specific area of internal walls is approximately assumed to be the same.

Secondly, the employed finite element model is based on static and linear elastic theory. This simulation model enables to present accurate physical behavior under small deformation. Because surgeons do not pinch the wall strongly in actual surgery, this hypothesis gives valid solution for satisfying haptic compatible calculation time.

The manipulators are simulated as two points using two PHANToMs. Although a glove-shaped device is desired in order to support complete motion of a human hand, the proposed approach performs simple implementation and efficiently enables pinching manipulation at a low cost.

The several experiments confirm that the developed system presents a discernible aorta model with normal stiffness, and sclerosis modeling enables grasping dangerous area through the virtual environment that simulates actual surgical procedures. However, some students cannot recognize the boundary through one use of the system. This result probably suggests recognizing boundary in virtual palpation is more difficult than to identify the approximate location of the sclerosis. According to the expert surgeons, this feature is the same as actual palpation. Because identifying the boundary is the next step after obtaining identification skill of normal and sclerosis stiffness, students may obtain the skill through repeated practice. Therefore, further examination is one future direction of this study in order to analyze whether the developed system enables such training. Also, further modeling of variable sclerosis cases are desired to verify the applicability as practical palpation simulator.

6. Conclusions

This paper presented a beating aorta model based on finite element method and provides a palpation simulator of aortas for preoperative rehearsal and training. The sclerosis aorta model was also proposed. 3D shape, stiffness and blood pressure can be applied to the model, which fulfills sufficient refresh rate for visual and force feedback in palpating the aorta model. Efficient pre-processing and fast computational scheme contributes to real time simulation.

Characteristic experiments with cardiovascular surgeons and medical students confirmed quality and performance of the developed system. The constructed aorta model with 1.0MPa efficiently simulates stiffness of in-vivo normal aortas. Another result demonstrates that the developed system is useful to recognize the normal stiffness of the virtual aorta model. The haptic feedback enables grasping sclerosis regions for rehearsal of surgical palpation. Further examination and modeling of other complex sclerosis cases are desired to increase and to verify the applicability of the developed system.

Acknowledgements

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