Real-time Simulation of Virtual Palpation System

Shan Liu, Yuhui Zhang, Wenfeng Zheng and Bo Yang

University of Electronic Science and Technology of China, Chengdu, Sichuan Province, China
ashanliu@uestc.edu.cn, bzhanguhui@uestc.edu.cn, cwinfirms@uestc.edu.cn, dboyang@uestc.edu.cn

Abstract. The current major difficulties are to establish accurate models to perform real-time simulation of soft tissue deformation for virtual palpation system, which is the contradiction between accuracy and real-time. This paper proposes a method that uses the parameters of the finite element method to estimate the elastic coefficient of mass-spring model and introduces the damping coefficient to rewrite the Euler integral, then builds a virtual palpation system for simulation. The experimental results show that the method is accurate while the real-time performance is guaranteed.

1. Introduction
Medical robotic system is one of the most active and fastest developing fields of robotics research in the world[1]. Palpation is a means of diagnosis and treatment of an unknown area of the organ tissue by the doctor. Adding force feedback to the medical robot to builds a virtual palpation system, can greatly increase the efficiency of doctors and improve the practical value of medical robotic system.

A virtual palpation system include soft tissue model, collision detection, deformation rendering and force feedback devices. The current research focuses on soft tissue modeling. Soft tissue model plays a critical role in the entire palpation system. Because of the unique material properties of the soft tissues, it is always difficult to establish a soft tissue model with high simulation. The current methods of soft tissue deformations can be divided into two categories: geometry-based and physics-based[2]. Physics-based model can better reflect the properties of soft tissue under external force. A variety of models have been produced, typically with Mass-Spring Model (MSM), Finite Element Model (FEM).

The mass-spring model appeared in the 1990s and was the first physical model proposed[3]. It discretizes the soft tissue into large number of mass points, which are connected by springs. The typical dynamic motion of the MSM model is described by Newton’s law. Significant research efforts have been dedicated to modelling of soft tissue deformation. Choi [4] uses a multi-layer mass-spring to establish a soft tissue deformation model, which effectively compensates for the defects of the single-layer model. Zhang Xiaorui [5] proposed a layered rhombus-chain-connection model. San-Vicente [6] discussed the mechanical properties of cubic MSM under tension and proposed a spring stiffness estimation method for both linear and nonlinear materials. There are also many improved mass-spring models [7,8,9,10]. Chanthasopeephan [11] reduced the dimension of the finite element model without reducing the accuracy of the feedback force. Khalaji [12] proposed a statistical finite element model. Tagawa [13] proposed a rectangular tetrahedral adaptive mesh based on synchronous rotating FEM. Cotin [14] compresses the linear matrix system produced by the volume finite element model to decrease computation time.

The mass-spring model has good real-time performance but poor accuracy, while the finite element
model has high simulation accuracy but the real-time performance is insufficient [15]. How to balance the accuracy and real-time of soft tissue deformation is an urgent problem to be solved. A good way is to combine the two models and use finite element method to estimate the elastic parameters of mass-spring, and to improve accuracy while ensuring real-time performance.

2. Model Formulation

2.1. Mass-Spring Model
The model commonly used to describe soft tissue behavior is Kelvin-Voigt mass-spring model [16]. When an external force is applied, the force of each influenced mass point is expressed as:

\[ M_i \ddot{X}_i + F_s_i + F_d_i = F_{ext}^{i} \]  

in which \( M_i, X_i, F_{ext} \) are mass, displacement vector and the external force at particle \( i \). \( F_s_i, F_d_i \) represent spring elastic force and damping force. They are described by:

\[ F_{s_i} = k_{s} (|x_{ij}^0 - x_{ij}|) \frac{x_{ij}}{|x_{ij}|} \]  

\[ F_{d_i} = k_{d} |v_{ij}| \frac{x_{ij}}{|x_{ij}|} \]

where \( k_s \) is the stiffness of springs, \( k_d \) is the damping coefficients. \( x_{ij}, v_{ij} \) and \( x_{ij}^0 \) are respectively the displacement, velocity vector and the initial length of the spring between particle \( i \) and \( j \). Their difference indicates the length change in the spring.

For the FEM, the kinetic equation of two particles can be given by:

\[ M_i \ddot{X}_i + A \sigma = F_{ext}^{i} \]  

\[ \sigma = E \epsilon \frac{|x_{ij}^0 - x_{ij}|}{x_{ij}^0} \frac{x_{ij}}{|x_{ij}|} \]

\( \sigma, \epsilon \) stands for the stress and strain of two particles, respectively. \( E \) is the Young’s Modulus of soft tissue. \( A \) is the cross-sectional area of FEM units.

2.2. Parameters
An important reason for the lack of mass-spring accuracy is that the selection of the parameters is purely empirical. So we use finite element method to estimate the elastic parameters of the mass-spring. Comparing Eq(1) and Eq(4), we can obtain:

\[ F_{s_i} + F_{d_i} = A \sigma \]

In most cases \( k_d \) is much smaller than \( k_s \), here we ignore the damping coefficient \( k_d \). By combining Eq(2) with Eq(5) we obtain Eq(7):

\[ k_s = \frac{AE}{x_{ij}^0} \]

According to the research results of previous[17,18,19], the Young’s Modulus \( E \) is not constant. Based on Eq(7), \( k_s \) has similar change trend with \( E \). The typical stress-strain curve is divided into two regions: linear region and nonlinear region. To better characterize soft tissue properties, we assume that the segmentation relationship between stress and strain is as follows:
\[
\sigma = \begin{cases} 
E \varepsilon, & \varepsilon \leq \varepsilon_0 \\
\alpha \varepsilon^2, & \varepsilon > \varepsilon_0 
\end{cases}
\] (8)

Plug Eq(8) to Eq(7):

\[
k = \frac{EA}{x_0} = A \frac{\partial \sigma}{\partial E} = \begin{cases} 
\frac{AE_0}{x_0}, & \varepsilon \leq \varepsilon_0 \\
A, & \varepsilon > \varepsilon_0 
\end{cases}
\] (9)

2.3. Integration

In order to perform real-time deformation simulation, it is necessary to dynamically calculate the displacement and velocity of the next moment. The Euler integral method is a commonly used numerical solution method for real-time simulation. The explicit Euler integral is unstable, especially for large steps. Implicit Euler integrals can solve the stability problem of explicit Euler integrals [20]. Since the damping coefficient is neglected in the estimation of the parameters, in order to better simulate the soft tissue characteristics, we reintroduce the damping coefficient \( k_d \) (0 < \( k_d \) < 1) to decelerate the velocity. At the same time, the calculation of force using implicit Euler integrals is difficult, so the explicit Euler integral is still used in calculation velocity. The final integral formula are as follows:

\[
\begin{align*}
\Delta v^{i+1} &= v^i + \Delta t \cdot \frac{F^i}{m} \\
\Delta v^{i+1} &= k_d \cdot \Delta v^{i+1} \\
\Delta x^{i+1} &= x^i + \frac{1}{2} \Delta t \cdot (v^{i+1} + v^i)
\end{align*}
\] (10)

The force \( F^i \) applied on particle i is computed at any time \( t \), where \( \Delta t \) is the iteration time-step defined in advance. The displacement and velocity at time \( t + \Delta t \) can be obtained with \( \Delta t = 0.01, k_d = 0.9 \).

2.4. Collision Detection

For realistic graphical display, collision detection is needed. Collision detection is determined between the force feedback device and the surface of the soft tissue model to find the nearest node to the collision point, and then an external force is applied to the particle. We calculate the displacement and velocity with the integral method mentioned above and update the location each time. Real time deformation is simulated.

3. Experiments

Experiments have been conducted to evaluate the performance of the proposed method. Performance evaluation focuses on soft tissue deformation effects and typical mechanical properties. We used the hardware platforms that include Geomagic Touch force feedback device which came from the 3D System Company, a computer with Intel(R) Core(TM) i7-4790 CPU @3.60GHz, 8GB RAM and acer LCD, as shown in Fig.1. All the experiments were developed with C++ and OpenGL on Visual Studio 2013. The virtual scene setting and force feedback effect came from the OpenHaptics.
Figure 1. The platform with a PC and a Geomagic Touch

The surface model of soft tissue is constructed using OpenGL. When the force $F$ is applied to the surface, the value of $F$ will changes with time $t$. After the stabilization, the magnitude of the applied force $F$ and the displacement data of the pressing portion are recorded. At present, the parameters of biological tissues are obtained from uniaxial indentation experiments[18][21]. In order to compare with the actual measured values, the data of our experiments is also performed in only one direction. That is, the Z-axis direction.

4. Results and Discussion

The display of the deformation process of a surface of soft tissue is shown in Fig.2. It contains 625 (25×25) nodes, and the deformation has relatively high reality. With the force-feedback devices, the force can be driven to the user. Fig.3 is the force-displacement behavior of the model drawn with MATLAB for the recorded data, while $E_a = 12000, A = 1e-4, a = 1500, \epsilon_r = 0.15$.

Figure 2. soft tissue deformation:no touch (left) and touch (right)

Figure 3. Comparison the force-displacement of the reference[21]
In reference [21], the author gives the maximum and minimum values of force-displacement data for real soft tissue (liver). As can be seen from the Fig.3, the force response of our model is within this range, which shows better accuracy. In terms of real-time, the refresh rate must be more than 25 frames per second [22]. The frame rate of our model is shown in Fig.4, it is basically more than 100 frame per second, fully able to meet real-time simulation and virtual display.

![Figure 4. Refresh rate of deformation and display](image)

5. Conclusion
In this paper, we use the finite element method to estimate the parameters of the spring-mass model, then introduce the damping coefficient to rewrite the Euler integral formula in the deformation simulation. In comparison with the previous MSM, our model has higher accuracy. Compared with the experimental data reported by reference, our model shows a certain accuracy, and the real-time simulation frame rate reaches more than one hundred, which can fully meet the real-time performance. Another contribution we make is that it provides a comprehensive framework for virtual palpation, from system design to implementation. In the future we should evaluate the effectiveness of the method on more complex models not only a surface.

6. Acknowledgments
This research was supported by the Fundamental Research Funds for the Central Universities (ZYGX2016J101). The authors would like to thank the anonymous reviewers for their invaluable suggestions and comments on the paper.

7. References
[1] Takacs A, TarJ K, Haidegger T, et al.: Applicability of the Maxwell-Kelvin model in soft tissue parameter estimation. IEEE, International Symposium on Intelligent Systems and Informatics, 115-119 (2015)
[2] Zhang J, Zhong Y, Smith J, et al.: ChainMail based neural dynamics modeling of soft tissue deformation for surgical simulation. Technology & Health Care Official Journal of the European Society for Engineering & Medicine (2017)
[3] Wang Z, Fratarcangeli M, Ruimi A, et al.: Real time simulation of inextensible surgical thread using a Kirchhoff rod model with force output for haptic feedback applications. International Journal of Solids & Structures, s 113–114:192-208 (2017)
[4] Choi K S, Soo S, Chung F L.: A virtual training simulator for learning cataract surgery with
phacoemulsification. Computers in Biology & Medicine, 39(11):1020-1031 (2009)

[5] Zhang X, Sun W, Song A.: Layered rhombus-chain-connected model for real-time haptic rendering. Artificial Intelligence Review, 41(1):49-65 (2014)

[6] San-Vicente G, Aguinaga I, Tomás C J.: Cubical Mass-Spring Model design based on a tensile deformation test and nonlinear material model. IEEE Transactions on Visualization & Computer Graphics, 18(2):228 (2012)

[7] Nikolaev, Sergei.: Non-linear mass-spring system for large soft tissue deformations modeling. Physics, P.88-94 (2014)

[8] Huangfu Z.: An Improved Mass-spring Model for Simulation of Soft Tissue Deformation. Journal of Information & Computational Science, 10(17):5551-5558 (2013)

[9] Basaá E, Farahmand F: Real-time simulation of the nonlinear visco-elastic deformations of soft tissues. International Journal of Computer Assisted Radiology & Surgery, 6(3):297-307 (2011)

[10] Wang S, Chu L, Fu Y, et al.: An Unfixed-elasticity Mass Spring Model based simulation for soft tissue deformation. IEEE International Conference on Mechatronics and Automation. IEEE, 309-314 (2014)

[11] Chanthasasopeeph T, Desai J P, Lau A C W.: Modeling Soft-Tissue Deformation Prior to Cutting for Surgical Simulation: Finite Element Analysis and Study of Cutting Parameters. IEEE transactions on bio-medical engineering, 54(3):349 (2007)

[12] Khalaji I.: Statistical finite element method for real-time tissue mechanics analysis. Comput Methods Biomech Biomed Engin, 15(6):595-608 (2012)

[13] Tagawa K, Yamada T, Tanaka H T.: A rectangular tetrahedral adaptive mesh based corotated finite element model for interactive soft tissue simulation. Engineering in Medicine & Biology Society. Conf Proc IEEE Eng Med Biol Soc, 7164 (2013)

[14] Bro-Nielsen M, Cotin S.: Real-time Volumetric Deformable Models for Surgery Simulation using Finite Elements and Condensation. Computer Graphics Forum, 15(3):57-66 (2010)

[15] Liu X, Wang R, Li Y, et al.: Deformation of Soft Tissue and Force Feedback Using the Smoothed Particle Hydrodynamics. Comput Math Methods Med, 2015(7):1-10 (2015)

[16] Takács Á, Rudas I J, Haidegger T.: Surface deformation and reaction force estimation of liver tissue based on a novel nonlinear mass-spring-damper viscoelastic model. Medical & Biological Engineering & Computing, 54(10):1-10 (2015)

[17] Omar N, Zhong Y, Smith J, et al.: Local deformation for soft tissue simulation. Bioengineered, 7(5):291-297 (2016)

[18] Yang J, Lingtao Y U, Wang L, et al.: STUDY ON MECHANICAL CHARACTERIZATION OF LIVER TISSUE BASED ON HAPTIC DEVICES FOR VIRTUAL SURGICAL SIMULATION. Journal of Mechanics in Medicine & Biology, 16(08):2169-2178 (2016)

[19] Rosen J, Brown J D, De S, et al.: Biomechanical properties of abdominal organs in vivo and postmortem under compression loads. Journal of Biomechanical Engineering, 130(2):021020 (2008)

[20] Duan Y, Huang W, Chang H, et al.: Volume Preserved Mass-Spring Model with Novel Constraints for Soft Tissue Deformation. IEEE Journal of Biomedical & Health Informatics, 20(1):268-280 (2015)

[21] Bao Y, Wu D, Yan Z, et al.: A New Hybrid Viscoelastic Soft Tissue Model based on Meshless Method for Haptic Surgical Simulation. Open Biomedical Engineering Journal, 7(1):116-124 (2013)

[22] Goulette F, Chen ZW.: Fast computation of soft tissue deformations in real-time simulation with hyper-elastic mass links. Comput Methods in Appl Mech Engg, 295:18-38 (2015)