A Constitutive Model of Soft Tissue Deformation for Virtual Surgical Simulation: A Literature Review

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Abstract. Realistic modeling of soft tissue deformation has been recognized as an essential requirement for virtual surgical simulators. As a consequence, the constitutive model is crucial for modeling soft tissue deformation. In this paper, the significance and characteristics of soft tissue deformation in virtual surgery simulation studies are introduced and analyzed its challenges. We have been focusing on investigating the constitutive relations in the soft tissue mechanical models: viscoelastic models and hyper-elastic models. Moreover, some techniques are given to determine and optimize model parameters through biomechanical experiments. Finally, based on the existing constitutive model, a novel notion for optimizing the soft tissue deformation constitutive model is proposed. The contribution of this paper is to provide theoretical guidance for the simulation of soft tissue deformation in the virtual surgery simulation system and to supply various practical suggestions for developing new constitutive models in the future.

Introduction

Virtual reality (VR) emerged as an essential technology of medical devices applications in recent years, benefiting from these cutting-edge facilities, surgical procedure simulations and robot-assisted surgery become increasingly immersive and reliable by adding visual and haptic feedback in the medical field [1]. Obtaining the interaction between the needle and the soft tissue is a crucial step to develop a surgical simulator with tactile sensation. The research and development of surgical simulation systems have potential applications for surgical training, practice and both pre-operative and intra-operative planning. It can also provide surgeons with a real-time visual and tactile information [2]. At present, with the continuous development of computer science and technology, biomechanics, graphics, and robotics, surgical simulations are mostly virtual augmented surgical simulation systems based on VR and Augmented Reality (AR) technologies. Virtual surgery simulation is one of the most universal applications of virtual reality in the medical field, which utilizes multi-model medical image data and virtual reality technology to generate a realistic surgical scene, thereby surgeons can use the virtual environment to simulate the surgical procedures. Surgical simulation system can simulate various physical and physiological characteristics of living tissue, it enhances the effectiveness of surgical training and reduce the risk of surgery [3]. In addition, it can provide a variety of different experimental objects for surgeons according to the specific needs, greatly cut down the surgeon's training costs compared with the animal or corpse-based training.

With the soft tissue is generally characterized by inhomogeneous, anisotropy, nonlinearity, viscoelasticity and hyper-elasticity, internal structure simulation is complex and sensitive to deformation. As a consequence, the study of soft tissue deformation in virtual surgery simulation system is the most important challenge at present, which directly determines the practicality and stability of the virtual surgery system. To simulate soft tissue deformation accurately in real-time, it is necessary to find a suitable constitutive model for soft tissue. The purpose of this paper is to provide a
literature review on previous constitutive models, moreover based on the existing constitutive models, a novel constitutive model was proposed for the development and optimization of constitutive models. The structure and content of this paper are divided into four parts: Firstly, the significance and characteristics of soft tissue deformation in surgical simulation are introduced briefly, then, we make a brief review of the constitutive relation of soft tissue: viscoelastic models and hyperelastic models. Thirdly, the parameters of constitutive model identification. Finally, give a summary of the overall study.

**Constitutive Models for Soft Tissue**

**Viscoelastic Models**

The mechanical behavior of biological tissues generally shows viscoelastic properties, the earliest research on viscoelastic mechanics can date back to the 19th century [4, 5]. The stress-strain relationship of viscoelastic materials is related to time, and the deformation of the viscoelastic materials is between elasticity and viscosity, its mechanical behavior mainly manifests as creep and relaxation properties. Linear viscoelasticity is mainly composed of two basic elements: a spring and a dashpot, and various viscoelastic models are generated through different elements combinations. In this paper, we just describe general viscoelastic models and their constitutive relations. Such as, in 1867, Maxwell proposed a simple linear viscoelastic model consisting of a spring and a dashpot in series, stiffness E represents spring-simulated elastic response [6], viscosity simulates the viscous response. Kelvin-Voigt model is made by connecting two basic elements in parallel [7]. The standard linear solid model is a viscoelastic material equivalent to a spring in series with a Kelvin model, it can describe the creep and relaxation properties of viscoelastic materials. The relationship be-tween stress and strain as shown in Table 1.

<table>
<thead>
<tr>
<th>Model</th>
<th>Constitutive structure</th>
<th>Constitutive equation</th>
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<tbody>
<tr>
<td>Maxwell</td>
<td>![Maxwell Diagram]</td>
<td>[\frac{d\sigma}{dt} + \frac{\sigma}{\tau} = E \frac{d\varepsilon}{dt}]</td>
</tr>
<tr>
<td>Kelvin-Voigt</td>
<td>![Kelvin-Voigt Diagram]</td>
<td>[\sigma = E\dot{\varepsilon} + \eta \frac{d\varepsilon}{dt}]</td>
</tr>
<tr>
<td>Standard linear solid model</td>
<td>![Standard Linear Solid Diagram]</td>
<td>[(E_1 + E_2)\sigma + \eta_1 \dot{\sigma} = E_1\dot{\varepsilon} + \eta_1E_2\dot{\varepsilon}]</td>
</tr>
<tr>
<td>Maxwell Standard linear solid model</td>
<td>![Maxwell Linear Solid Diagram]</td>
<td>[(E_1 + E_2)\dot{\sigma} + E_2\frac{\sigma}{\tau_1} = \dot{\varepsilon} + \frac{\sigma}{\tau_1}]</td>
</tr>
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where, \(\sigma = \text{stress}, \varepsilon = \text{strain}, \tau = \eta \frac{1}{E}, \tau_1 = \eta_1 \frac{1}{E_1}\) is relaxation time, \(\dot{\sigma} = \frac{d\sigma}{dt}, \dot{\varepsilon} = \frac{d\varepsilon}{dt}\).

Due to space limitations, this chapter only introduces common viscoelastic models, and many viscoelastic models are often used. For example Hunt-Crossley model, Hybrid model, Boltzman model and Generalized Fractional Derivative Model [8, 9, 10]. In 1993, the four-parameter model proposed by [11]. Many scholars have been researched on linear viscoelasticity, but as we know that simple homogeneity and isotropic linear elasticity model is difficult to describe the realistic mechanical behavior of soft tissue organ due to the complex structure. Thus, it is increasingly important to develop nonlinear viscoelastic models.
Hyperelastic Models

The concept of pseudo-elasticity for soft tissue was proposed by [12]. Fung states that the techniques of elasticity theory can be borrowed to deal with an inelastic material. The most often used the concept of pseudo-elasticity is the assumption that a strain energy function exists, that is hyperelasticity. Although an approximation, it is very convenient to utilize the strain energy function for soft tissue modeling. To find a realistic strain energy would require characterization of soft tissue in various deformation modes [13]. Some general hyper-elastic models detailed description in [14, 15, 16, 17, 18, 19].

Once the strain energy function $W$ is determined, using the relationship between the Kirchoff stress tensor $\sigma_{ij}$ and the Green strain tensor $\varepsilon_{ij}$ to generate constitutive equation, it is given by:

$$\sigma_{ij} = \frac{\partial W}{\partial I_1} \frac{\partial I_1}{\partial \varepsilon_{ij}} + \frac{\partial W}{\partial I_2} \frac{\partial I_2}{\partial \varepsilon_{ij}} + \frac{\partial W}{\partial I_3} \frac{\partial I_3}{\partial \varepsilon_{ij}}$$  \hspace{1cm} (1)

The relationship between the principal stress and its principal stretches is obtained:

$$\sigma_{ij} = 2\left(\lambda_i^2 \frac{\partial W}{\partial I_1} + \lambda_i^{-2} \frac{\partial W}{\partial I_2}\right) + P$$  \hspace{1cm} (2)

where, $I_1$, $I_2$ denote the first and second strain invariants of the deviatoric strain, respectively, $\lambda_i$ (i=1,2,3) the principal stretches, $P$ represents an unknown hydrostatic pressure, which is determined by the stress [20].

Model Parameters Identification

The constitutive models have been discussed in the above mentioned analysis, the next step is how to determine the material parameters in the constitutive model. The method generally used is to obtain the real material data of soft tissue through biomechanical experiments [21, 22, 23, 24], mainly stress-strain data. In addition, the elastic parameters of the material can also be measured, such as Young's modulus, Shear modulus and Poisson ratio. Then, using the experimental data to fit the material parameters in a constitutive model, there is some optimization algorithm fit the data well. Finally, finite element (FE) commercial software is utilized to verify the reliability of the model through simulation. With the combination of experimental data and FE software, the parameters are adjusted and optimized by comparing the experimental results with the analysis simulation results [25, 26, 27, 28]. The finite element (FE) commercial software includes ABAQUS, ANSYS, COMSOL and so on. Uniaxial tension, uniaxial compression, share and indentation are often used for biomechanical experiment, especially, the experiment of tension and compression, as shown in Figure 1. The parameters of the material are determined by the least square approach, for a given set of stress and strain experimental data, the minimum value of relative error $E$ is given by:

$$E = \sum_{i=1}^{n} \left(1 - \frac{T_{i}^{th}}{T_{i}^{ex}}\right)^2$$  \hspace{1cm} (3)

where, $T_{i}^{th}$ = the stress value in the experimental data, $T_{i}^{ex}$ = the stress value corresponding to the principal stretch ratio in the constitutive relation.
In recent years, increasing number of researches have been focused on simulating soft tissue deformation by using FE software. Real-time finite element methods have been adapted for the simulation of nonlinear constitutive models with the inclusion of haptic feedback [29, 30]. The soft tissue deformation was simulated under indentation, the Arruda–Boyce equation was utilized as a behavior model of tissue characteristics [31]. Tendick et al. using Mooney-Rivlin FEA Model to simulate tissue deformation in training environment [32]. An endoscopic simulator has been developed utilizing a Neo-Hookean material law for the surrounding tissue [33]. Further, many scholars combined the technology of finite element methodology with constitutive model and described the mechanical behavior of soft tissue well during simulation. More complex nonlinear constitutive models and biomechanical experiments need to find and design, an experimental device obtaining force-displacement information was presented by [34], utilizing a constitutive model developed from ex vivo experimentation for the simulation of in vivo tissue loading was tested, and the results showed the accuracy of the force response was improved, as we can see in Figure 2(a). Finite element method (FEM) is a common method in modeling soft tissue with its accuracy, a finite element and meshless modeling method was used to simulate the deformation of a large intra-abdominal organ during laparoscopic surgery, the results indicated that meshless modeling can provide reasonable accuracy with low computational cost [35], as shown in Figure 2(b). Calibrating mod-el by using a least-squares method is an effective way to develop an accurate material model. Based on the uniaxial test data, the stress-strain curves were generated and corresponding model was built in ABAQUS, using the least-squares method to calibrate exponential model and incorporated it into a nonlinear finite element framework to simulate the behavior of soft tissue under a large-scale deformation [36], that shows by using the calibrated exponential model and nonlinear FEM in a simulation can achieve real-time response under large deformation, as shown in Figure 2(c). To above researches clearly show that combination between FEM and constitutive model is a promising approach for predicting soft tissue behavior and identifying the soft tissue properties.
Figure 2. (a) The results of soft tissue deformation simulation and comparison of the experimental and simulated probing force [34]. (b) The meshless method for modeling kidney [35]. (c) A large-scale deformation effect using the exponential material model [36].

Summary

Soft tissue deformation is a crucial part of the virtual surgery simulation system, due to various kinds of soft tissues, with different mechanical properties, we focus the biomechanical properties of soft tissue in this paper. Deformable human soft tissue simulation with real-time and accuracy requirements in virtual surgery is needing to find a precise and effective constitutive model. Compared with the linear viscoelastic model, the nonlinear viscoelastic model demonstrates an excellent performance on the properties of the real soft tissue material, yet it also shows challenges in developing a nonlinear viscoelastic model considering the characteristics of the material and the loading conditions. A hyper-elastic model could handle the description of soft tissue stable with large deformable situations. In the future, to simulate the deformation of soft tissues more accurately, researchers need to combine viscoelasticity and hyperelasticity to establish a hyper-viscoelastic constitutive model. Combination the technology of FEM with a constitutive model will be a potential approach for simulating the soft tissue deformation in virtual surgery simulation system. Moreover, more biomechanical experiments require to be conducted to determine and optimize the model parameters.

References


