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Finite Element Analysis for evaluating liver tissue damage due to mechanical compression

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ABSTRACT

The development of robotic-assisted minimally invasive surgery (RMIS) has resulted in increased research to improve surgeon training, proficiency and patient safety. Minimizing tissue damage is an essential consideration in RMIS. Various studies have reported the quantified tissue damage resulting from mechanical compression; however, most of them require bench work analysis, which limits their application in clinical conditions of RMIS. We present a new methodology based on nonlinear finite element (FE) analysis that can predict damage degree inside tissue. The effects of the boundary conditions and material property of the FE model on the simulated von Mises stress value and tissue damage were investigated. Four FE models were analyzed: two-dimensional (2D) plane strain model, 2D plane stress model, full three-dimensional (3D) model, and 3D thin membrane model. Nonlinear material properties of liver tissue used in the FEA were derived from previously reported in vivo and in vitro experiments.

Our study showed that for integrated von Mises stress and tissue damage computations, the 3D thin membrane model yielded results closest to the full 3D analysis and required only 0.2% of the compute time. The results from 3D thin membrane and the full 3D models fell below plane-strain model and above the plane-stress model. Both stress and necrosis distributions were impacted by the material property of FE models. This study can guide engineers to design surgical instruments to improve patient safety. Additionally it is useful for improving the surgical simulator performance by reflecting more realistic tissue material property and displaying tissue damage severity.

1. Introduction

Extensive research on robotic-assisted minimally invasive surgery (RMIS) has focused on haptic feedback during grasping (Hu et al., 2004; MacFarlane et al., 1999; Rosen et al., 1999; Tholey et al., 2005; Vakili et al., 2011), however, little attention has been given to tissue damage resulting from grasping. A number of previous studies have quantified tissue damage specifically from general mechanical compression. Famaey et al. (2010, 2013) quantitatively studied the damage to smooth muscle cells of rat abdominal arteries after in vivo clamping to well-defined loading levels by using an isometric contraction model. De et al. (2007)and De (2008) proposed a damage–stress relationship based on measuring hepatic necrosis as a function of stress from in vivo experiments. Our study adopted De et al.'s quantitative study of necrosis for two reasons. First, we have extensive experimental data from De, 2008 and Rosen et al.'s (2008) earlier works. Second, the liver is widely modeled as isotropic and homogeneous which simplifies parameter identification and FEM.

The results from the above quantitative studies can be used by finite element analysis (FEA) to predict tissue damage in RMIS simulator. However, a tissue model needs to be established and validated for such surgery simulation (Fu and Chui, 2014). It is well known that most soft tissue is nonlinear, inhomogeneous and viscoeleastic (Fung, 1981). However, liver is considered to be a good model for continuum mechanics study because it is approximately isotropic and homogenous in uniform histology cross sections (Carter et al., 2001; Chui et al., 2004). If the strain rate effect is excluded, that is to say, the strain only changes with location but not with time, the mechanical response of liver tissue can be described by elastic model (Gao et al., 2010).

Uniaxial tension and compression are commonly used methods to investigate isotropic and homogeneous material. There are numerous published studies covering uniaxial tests on various soft tissues (Fu et al., 2014; Chui et al., 2004; Yarpuzlu et al., 2013), however, most of them were performed in vitro. The research

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presented in Rosen et al. (2008) is the only published study in vivo compression testes using a motorized endoscope grasper.

Our study investigates whether there is significant difference in simulating tissue damage between various in vivo and in vitro experiments. Additionally, we compare and contrast the effectiveness of the two-dimensional (2D) and three-dimensional (3D) FEA simulation in investigating the mechanical response of soft tissue to external loads. 2D models are preferred in industry because they are quicker and simplify the analysis during the initial design phase (Krueger et al., 2002; Romeed et al., 2006). However, there could be some physical characteristics that exist in 3D models that cannot be simulated in a 2D model. At this time, few reports have evaluated how these geometrical assumptions affect the reaction force and tissue damage computations in RMIS simulation.

The objective of the current study is to investigate the effects of the boundary conditions and material properties of the FE liver tissue model on the damage prediction for compression in RMIS. This type of investigation requires a balance between real world and computational efficiency. In this study, we extended earlier 2D non-linear finite element analyses to 3D, dramatically increasing the size of the computation. The liver tissue was considered nonlinear to maximize accuracy and homogenous and elastic to increase computational efficiency. Nonlinear finite element analyses were performed using ANSYS Mechanical APDL 14.0 under various geometrical assumptions: 2D plane strain, 2D plane stress, full 3D, and 3D thin membrane. The results of stress distribution and the integrated stress value were compared among the above four models. Tissue damage was calculated from the stress distributions based on De, 2008 necrosis–stress function. Additionally, three nonlinear material properties were studied in the full 3D model. Finally, the computed stress distribution, integrated stress value, as well as the tissue damage in the form of hepatic necrosis were compared.

2. Methods

2.1. Material model for tissue and grasper

The modeling methods for this research are similar to a recent study (Cheng and Hannaford, 2014) in which 2DOF simulations were used. Soft tissue is often inhomogeneous and anisotropic and its compounds vary throughout the whole structure. Carter et al. (2001) found solid organs, particularly the liver and spleen, could be treated as approximately isotropic and homogenous due to their uniform histology across species. Additionally because of the high water content of the liver tissue, we can consider it to be incompressible (Gao et al., 2010).

We adopted the nonlinear biomechanical property of the liver from Rosen et al.'s (2008) experiments. The porcine liver was tested both in vivo and in vitro under uniaxial compressive loadings using a custom-made device called the Motorized Endoscopic Grasper (MEG) and in vitro using a servohydraulic material testing system by MTS Corporation (Eden Prairie, MN). Compression stress (σ) and strain (ε) experimental data were plotted with the associated elastic phenomenological curve fitting function was represented as (Rosen et al., 2008) follows:

$$\sigma = \beta(e^{\alpha\varepsilon 2} - 1) + \gamma\varepsilon \tag{1}$$

Another common approach for studying the mechanical behavior of soft tissue is to use the constitutive physical-law based model (Fu et al., 2014; Rosen et al., 2008). This study employs the Ogden model, which is a well-established hyperelastic material model to describe the nonlinear stress–strain material behavior (Ogden, 1972). In Ogden model, the material behavior can be described by means of the strain energy density function

$$W_O = \sum_{i=1}^{N} \frac{\mu_i}{\alpha_i}\left[\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 + \frac{1}{d_i}(J-1)^{2i}\right] \tag{2}$$

where N is the model's order, α_1 and μ_1 are the parameters to be determined experimentally, and d_i represents the change in volume. λ_k (k=1–3) and J refer to the principal stretch and the Jacobian of deformation gradient respectively. Assuming that the liver is incompressible, the Jacobian of the deformation gradient becomes 1 ($J = \lambda_1\lambda_2\lambda_3 = 1$). Eq. (2) can then be simplified as below

$$W_O = \sum_{i=1}^{N} \frac{\mu_i}{\alpha_i}[\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3] \tag{3}$$

Classical continuum mechanics provides us with he principal stress, $\sigma_i = \lambda_i(\partial W_O/\partial\lambda_i)$, where λ_1 represents the stretch ratio in the direction of compression and σ_1 represents the corresponding principal stress. Since approximately an unconfined compression test was performed in Rosen et al.'s (2008) study, the other two principal stresses are approximated as zero ($\sigma_2 = \sigma_3 = 0$). Thus we will have $\lambda_2 = \lambda_3 = \lambda_1^{-1/2}$. Plugging λ_2 and λ_3 into Eq. (3) yields

$$\sigma_c = \sum_{i=1}^{N} \mu_i[\lambda_1^{\alpha_i} - \lambda_1^{-\alpha_i/2}] \tag{4}$$

Adopting the notation of Rosen et al. (2008), with the compression load, $\sigma_c = -\sigma$, and $\lambda = 1-\varepsilon$, we can then numerically fit Rosen et al.'s phenomenological stress–strain curves to Eq. (4) by the following steps: (1) The phenomenological model parameters β, α and γ in Rosen et al.'s (2008) study were the statistical mean values derived from several measures. For each measure, the liver tissue failed at different maximum strain, ranged from 0.35 to 0.6. Thus we choose strain ranged from 0 to 0.35 as the data-fitting region. (2) The corresponding stress is calculated by Eq. (1) by the increments of 0.01. (3) The strain and stress from step 2 are imported into MATLAB Curve Fitting Tool (Mathworks Inc.). Using the trust-region algorithm and least absolute residuals (LAR) method, we can obtain the parameters in 1st-order Ogden model with the coefficient of determination $R^2 < 0.98$ (see Table 1).

The plot in Fig. 1 compares the stress–stretch curves of the Ogden fit model with the original phenomenological curves in Rosen et al.'s study. In the region where 1 > stretch > 0.8, Fig. 1 shows that the MEG in vivo curves are close to MEG in vitro curves but far from MTS in vitro curves.

Surgical graspers are considered to be composed of an isotropic linear elastic material (stainless steel) with the Young's modulus E=190 GPa and the Poisson's ratio ν=0.27 (De et al., 2007).

2.2. Two-dimensional finite element models

Two assumptions are made for the 2D models: the plane strain model, which assumes the out of plane-strains are zero and the plane stress model, which assumes the out of plane-stresses are zero.

The outline of 2D models of the tissue–grasper contact shows the geometry and applied displacement vector (Fig. 2A). The dimension of the liver tissue slice is set to 10 mm height by 20 mm width. The grasper is set to be 2 mm height by 5 mm width. Because of the symmetry during grasping, only half of the tissue (5 mm height × 20 mm width) is considered in the analysis. The nodes on the bottom line (y = −5 mm) can only have freedom along the x direction. The applied

Table 1
The parameters for the least-square fitting 1st-order Ogden model with the phenomenological model under various experiment conditions.

Experiment condition	Phenomenological model (Ogden, 1972)	1st-order Ogden model
MEG, in vivo	β=7377, α=21, γ=3289	α_1=16.02, μ_1=0.002934
MEG, in vitro	β=7972, α=20, γ=781	α_1=15.44, μ_1=0.003163
MTS, in vitro	β=8450, α=26, γ=1679	α_1=20.82, μ_1=0.002227

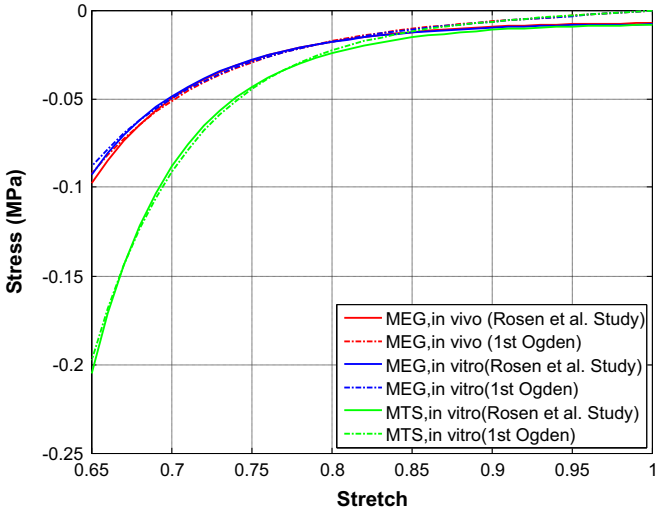

Fig. 1. Stretch–stress curves for various nonlinear tissue models.