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

Lei Cheng, Blake Hannaford

Department of Mechanical Engineering, University of Washington, Box 352600, Seattle, WA 98195, United States
Department of Electrical Engineering, University of Washington, Box 352500, Seattle, WA 98195, United States

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.

© 2015 Elsevier Ltd. All rights reserved.

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 viscoelastic (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
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 2D finite element 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 ($\sigma$) and strain ($\epsilon$) experimental data were plotted with the associated phenomenological curve fitting function was represented as (Rosen et al., 2008) follows:

$$\sigma = \beta(\epsilon^2 - 1) + \gamma \epsilon$$  (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

$$W_0 = \sum_{i=1}^{N} \mu_i \alpha_i \left[ \lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 \right]$$  (2)

where $N$ is the model’s order, $\alpha_i$ and $\mu_i$ are the parameters to be determined experimentally, and $\lambda_i$ represents the change in volume. $\lambda_i$ 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 $J = \lambda_1 \lambda_2 \lambda_3 = 1$. Eq. (2) can then be simplified as below

$$W_0 = \sum_{i=1}^{N} \mu_i \alpha_i \left[ \lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 \right]$$  (3)

Classical continuum mechanics provides us with he principal stress, $\sigma_i = \lambda_i (\partial W_0/\partial \lambda_i)$, where $\lambda_i$ represents the stretch ratio in the direction of compression and $\sigma_i$ 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 $\lambda_2$ and $\lambda_1$ into Eq. (3) yields

$$\sigma_i = \sum_{i=1}^{N} \mu_i \alpha_i \left[ \lambda_1^{\alpha_i} - 1/2 \right]$$  (4)

Adopting the notation of Rosen et al. (2008), with the compression load, $\sigma_i = -\sigma$, and $\lambda_i = 1 - \epsilon$, 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 $\beta$, $\alpha$ and $\gamma$ 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 $R^2 < 0.08$ (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 $\epsilon < 0.1$ and $\sigma < 190$ GPa and the Poisson’s ratio $\nu = 0.27$ (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 x 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>
<thead>
<tr>
<th>Experiment condition</th>
<th>Phenomenological model</th>
<th>1st-order Ogden model</th>
</tr>
</thead>
<tbody>
<tr>
<td>MEG, in vivo</td>
<td>$\beta = 7377$, $\alpha = 21$, $\gamma = 1289$</td>
<td>$\alpha_1 = 16.02$, $\mu_1 = 0.002934$</td>
</tr>
<tr>
<td>MEG, in vitro</td>
<td>$\beta = 7972$, $\alpha = 20$, $\gamma = 781$</td>
<td>$\alpha_1 = 15.44$, $\mu_1 = 0.003163$</td>
</tr>
<tr>
<td>MTS, in vitro</td>
<td>$\beta = 8450$, $\alpha = 26$, $\gamma = 1679$</td>
<td>$\alpha_1 = 20.82$, $\mu_1 = 0.002227$</td>
</tr>
</tbody>
</table>

Fig. 1. Stretch–stress curves for various nonlinear tissue models.
displacement of the grasper increases from 0 mm to 2 mm by the increments of 0.01 mm in the negative y direction \( (U_y = 0 \text{ to } -2 \text{ mm}) \).

The tissue is modeled with 4-noded axisymmetric, quadrilateral, solid elements. The meshing size for the tissue is uniform to facilitate the comparison with three-dimensional models. The grasper domain is meshed using the same element type but its size is gradually refined for the portions which will potentially be involved in the grasper–tissue interaction (Fig. 2B).

### 2.3. Three-dimensional finite element models

The three-dimensional (3D) model of the tissue and grasper with geometry and displacement is shown in Fig. 3A. Two 3D models are analyzed in the study: the 3D thin membrane model and the full 3D model. The dimensions of the thin membrane model (Fig. 3B) are: 10 mm high, 20 mm wide, and 0.2 mm thick. The size of grasper is set at 2 mm high, 5 mm wide and 3 mm thick. For the full 3D model, dimensions of the tissue and grasper blocks are 10 mm high, 20 mm wide, 20 mm thick and 2 mm high, 5 mm wide, 3 mm thick respectively (Fig. 3C).

Similarly to the 2D models, because of the symmetry during grasping, only half of the tissue is considered in our analysis. The nodes on the bottom surface \( (y = -5 \text{ mm}) \) can only have freedom along the x and z direction. The applied displacement of the grasper increases from 0 mm to 2 mm by the increments of 0.01 mm in the negative y direction \( (U_y = 0 \text{ to } -2 \text{ mm}) \). In the 3D model the meshing size for the grasper is set to be uniform at 0.5 mm. Table 2 lists the meshing size and element type of tissue for each model.

To compare the results of 3D with the 2D models, the center cut plane along the z direction is extracted from the 3D simulation results. Fig. 3E illustrates the

---

**Fig. 2.** 2D Finite element model of tissue and grasper. (A) Illustration of geometry and applied displacement. (B) meshing size of tissue = 0.05 mm.

**Fig. 3.** 3D Finite element model of tissue and grasper. (A) Geometry and applied displacement in full 3D model. (B) 3D thin membrane model. (C) Full 3D model. (D) Refined meshing of tissue for full 3D model. (E) Center cut plane (blue area) in full 3D model. Red areas sketch tissue while gray solid areas sketch grasper. (A) 2D Plane strain. Meshing size = 0.05 mm. (B) 2D Plane stress. Meshing size = 0.05 mm. (C) 3D thin membrane. Cut plane. Meshing size = 0.05 mm. (D) Full 3D. Cut plane. Meshing size = 0.2 mm. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)
2.4. Calculation of tissue damage

We utilized the stress magnitude value from De (2008) study. In three hours after animals were euthanized, the test tissues were harvested and processed for staining. Multiple sections were taken from each damaged tissue sample parallel to the direction of compression. Cell death, as represented by percent area of necrosis, was quantified by image analysis software. Through a nonlinear mixed effect model, the necrosis as a function of stress was obtained as below (De, 2008)

\[
\% \text{Necrosis} = a \frac{B \sigma_{\text{stress}}^C}{D + \sigma_{\text{stress}}^C} \times \text{Duration} \times E
\]

(5)

\( \sigma_{\text{stress}} \) and Duration refer to equilibrium von Mises stress and duration of grasping (seconds) respectively. It should be noted that since the value of \( a \) is negative (\( a = -4.27 \)) in Eq. (5), the general population fit would show negative necrosis at zero stress, which is biologically impossible. De (2008) held the compression stresses for Duration at 10, 30 and 60 s and found the factor of duration had a very small effect. Thus Duration is not considered in our study and Eq. (5) can be simplified as

\[
\% \text{Necrosis} = a \frac{B \sigma_{\text{stress}}^C}{D + \sigma_{\text{stress}}^C}
\]

(6)

The parameter estimates for the best fit nonlinear mixed effect model are \( B = 81.42 \), \( C = 4.67 \) and \( D = 176.92 \) (De, 2008).

In 2D, the total equivalent tissue damage area \( D_{\text{damage}} \) in the entire area (5 mm high \( \times \) 20 mm wide) is calculated in the following function:

\[
D_{\text{damage}} = \sum_i \% \text{Necrosis}_i \times A_i
\]

(7)

where \( \% \text{Necrosis}_i \) refers to percent of necrosis at the \( i \)-th point. \( A_i \) refers to the contributory area of the \( i \)-th point. In 2D, the sum of the \( A_i \) is equal to 5 mm high \( \times \) 20 mm wide = 100 mm². In 3D, the total equivalent tissue damage volume is 2000 mm³ (5 mm high \( \times \) 20 mm wide \( \times \) 20 mm thick) and damage volume is calculated by the following equation:

\[
D_{\text{damage}} = \sum_i \% \text{Necrosis}_i \times V_i
\]

(8)

where \( V_i \) refers to the contributory volume of the \( i \)-th point.

In order to investigate the relationship between total tissue damage and total stress, the integrated von Mises stress, \( \text{Stress}_{\text{sum}} \), in 2D space and Stressy, in 3D space, is calculated similarly

\[
\text{Stress}_{\text{sum}} = \sum_i \sigma_{\text{stress}}
\]

(9)

\[
\text{Stress}_y = \sum_i \sigma_{\text{stress}} \times V_i
\]

(10)

\( \sigma_{\text{stress}} \) refers to the von Mises stress at the the \( i \)-th node.

3. Results

Four examples of the calculated von Mises stress distribution under compression displacement with various boundary conditions are given in Fig. 4A–D. For the 2D plane strain model, 80% of the stress in the area directly beneath the grasper was over 300 kPa (Fig. 4A). For 2D plane stress, most of the stress in the area directly beneath the grasper was around 80 kPa (Fig. 4B). For both the 3D thin membrane cut plane and the full 3D cut plane, the stress on the contact surface peaked at the two grasper corners to over 300 kPa, and reduced in the middle. Directly beneath the grasper, both 2D models had stress value around 200 kPa.

After mapping stress to percentage necrosis of each element area using Eq. (5), we created a computed necrosis distributions in 3D space. One plotted example is shown in Fig. 5, which indicates that the stresses around the corners and edges at the grasper–tissue contact surface contributed most to the tissue damage (red color). The tissue damage decreased with the increasing distance from the contact surface, but remained localized in XZ plane to the region of contact.

The integrated von Mises stress was calculated through adding up each element's von Mises stress multiplied by its contributory area (Eq. (9) and Fig. 6A). The tissue damage in the form of necrosis percentage was calculated using Eq. (5) (Fig. 6B). For each displacement shown in Fig. 6A and B, both integrated von Mises stresses and necrosis percentage of 3D thin membrane model were closest to those of full 3D model, with peak differences of 6% and 17% respectively. However, the computational time for 3D thin membrane was only 0.2% of the time for the full 3D model. Both the 3D thin membrane and the full 3D model yielded results that fell between 2D plane strain (upper limit) and 2D plain stress (lower limit). Due to the nonlinear property of the necrosis–stress curve, the ratio of integrated von Mises stress for the full 3D model over 2D plane model was different from that of the necrosis percentage. For example, at the displacement \( U_y = -1.8 \) mm, the 2D plane strain model yielded integrated von Mises stress 0.4 times more than the full 3D model while its calculated necrosis percentage was 1.1 times more than that of full 3D model.

A histogram showing the element percentage for three levels of von Mises Stress (Fig. 7A and B) was calculated for different nonlinear material properties (Table 1). Fig. 7A showed that in full 3D space, the results of the material properties derived from MEG in vivo and MEG in vitro had the similar stress distribution in all three ranges. The material property derived from MTS in vitro yielded less percentage of stress at lower compression (higher stretch) range, but higher percentage of stress at higher compression (lower stretch) range. Similar trends were seen in 3D cut plane space (Fig. 7B). The volume percentages of liver tissue damage in full 3D space for these three material properties were also calculated (Fig. 8). The results indicated similar tissue damage with MEG in vivo and MEG in vitro; also for each
displacement step, material properties derived from MTS in vitro yielded more tissue damage than both MEG in vivo and MEG in vitro. The integrated von Mises stress and predicted tissue damage area were calculated for uniform and refined meshing size of liver tissue. The results indicated that both integrated force and damage were slightly impacted by the two meshing methods. However, for each displacement step, the calculation time was decreased by 72–85% with the refined meshing size method.

4. Discussion and conclusions

As far as we know, our study is the first to match continuum mechanics model with in vivo uniaxial experimental data of liver. Also it is the first to use nonlinear tissue property to predict tissue damage resulting from grasping in RMIS. This study can be useful for researchers to develop and test various design of graspers for improving patient safety and surgeon training. Instruments which can automatically display the tissue damage may decrease surgical complications. The surgical simulator performance can be improved by reflecting more realistic tissue material property and predicting tissue damage for the student.

The influence of the boundary and material assumptions made in developing liver tissue damage model on grasper–tissue interaction was studied. Geometrically nonlinear finite element analyses using 2D plane strain, 2D plane stress, 3D thin membrane and full 3D model were performed. Three different nonlinear material properties were derived from published in vivo and in vitro studies. The von Mises stress distribution, integrated von Mises stress and equivalent tissue damage percentage were compared for various geometrical boundaries and material properties. In order to investigate the meshing sensitivity of the FE model, the results from two meshing methods were recorded. Our results showed that results from the 2D plane strain and 2D plane stress models formed the upper and lower bound of the results obtained from full 3D model cut plane. The results from the 3D thin membrane model were very close to the results from the full 3D model in cut plane (Fig. 6).

We found that both von Mises stress distribution and tissue damage were impacted by the nonlinear material property values in our data set. The results from MEG in vivo and MEG in vitro are close but there exist obvious differences between them and the results from MTS in vitro. With the MTS parameters, computations yielded more area with higher stress. As a result, the calculated tissue damage percentage was much higher for MTS in vitro. The significant differences between MEG and MTS results may be caused by the large degree of variability between MEG and MTS tests, such as differences of loading controls and boundary conditions. While this variability makes finding a universal numerical model of tissue damage difficult, it is worthwhile to check the consistence of the experimental conditions of stress-damage tests with those of strain–stress tests for a more accurate prediction of liver tissue damage.

3D models clearly represent allow for more accurate representation of practical problems. However, full 3D FEA consumes much more modeling and computational time than 2D FEA. Our study indicated that a 3D thin membrane model may be a good alternative to full 3D
model since it can yield the closest results to full 3D model while dramatically reducing the computational time by 99.8%. It was also found that refining meshing can reduce computational time by over 70% and have little impact on the results.

4.1. Limitations and future directions

Our refined meshing tissue model calculation was not able to converge if the applied displacement was larger than 1.4 mm, i.e. 28% vertical applied strain. Some approaches we tried to overcome this computational difficulty included curving the edge of the grasper, adjusting the computational parameters normal penalty stiffness (FKN) and penetration tolerance (FTOLN), decreasing the sub-step of displacement, and applying other algorithms in ANSYS, but model convergence beyond 28% strain was not obtained.

The liver tissue material in this study was assumed to be isotropic and homogenous. Additionally the tissue’s mechanical properties were assumed not to change after the tissue cell was damaged. In reality, the liver tissue is biologically complex and is composed of various substructures like stroma and hepatocytes with different stress-strain characteristics. After being damaged, the crushed vessels inside the tissue can cause blood loss and the dead tissue cells may release a number of chemicals, both of which would change the local material properties.

The present study discussed the stress concentrations and related the stress results to the tissue damage based on the tissue damage-stress function as studied by De (2008). However, there may be different damage mechanisms. Recent research proposed excessive strain as another direct factor for initiating pressure-induced deep tissue injury (Breuls et al., 2003; Ceelen, et al., 2008). Future studies should also compare the strain with stress concentration and which factor has a stronger correlation with the measured tissue damage.

With additional in-vivo validation studies, we hope to develop an accurate FE model with more feasible computational time, by which

Fig. 6. Integrated von Mises stress and tissue damage area percentage in 2D space with nonlinear tissue properties derived from MEG in vivo experiments. (A) Histogram of stresses in 3D space. (B) Histogram of stresses in 3D cut plane space.
the tissue damage resulting from mechanical compression in RMIS can be displayed to surgeons in real-time. Also it could improve the surgical simulator performance for the student or professional. Researchers can use the 3D computational models to rank different designs of grasper performance while causing the least damage, or they can use a computationally efficient 2D model for early warning during a procedure. In a relative measurement, the absolute accuracy is less important than relative rankings within each model. A computationally efficient model may be used with force sensing and possibly machine vision (segmentation and simple measurements) to provide an early warning system for liver damage during robotic surgery.

**Conflicts of interest statement**

The authors whose names are listed immediately below certify that they have NO affiliations with or involvement in any organization or entity with any financial interest (such as honoraria; educational grants; participation in speakers' bureaus; membership, employment,
consultancies, stock ownership, or other equity interest; and expert testimony or patent-licensing arrangements), or non-financial interest (such as personal or professional relationships, affiliations, knowledge or beliefs) in the subject matter or materials discussed in this manuscript.

Acknowledgment

This research was funded by NSF (60167194). We would like to acknowledge Dr. Eric Seibel, Department of Mechanical Engineering, University of Washington for his comments and advices in this study. Special thanks to Sophia Hannaford for editing.

References