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A multi-physics model for ultrasonically activated soft tissue

Rahul Suvranu De*

Department of Mechanical, Aerospace, and Nuclear Engineering, Rensselaer Polytechnic Institute, 110 8th Street, Troy, NY 12180 USA

Abstract

A multi-physics model has been developed to investigate the effects of cellular level mechanisms on the thermomechanical response of ultrasonically activated soft tissue. Cellular level cavitation effects have been incorporated in the tissue level continuum model to accurately determine the thermodynamic states such as temperature and pressure. A viscoelastic material model is assumed for the macromechanical response of the tissue. The cavitation model based equation-of-state provides the additional pressure arising from evaporation of intracellular and cellular water by absorbing heat due to structural and viscoelastic heating in the tissue, and temperature to the continuum level thermomechanical model. The thermomechanical response of soft tissue is studied for the operational range of frequencies of oscillations and applied loads for typical ultrasonically activated surgical instruments. The model is shown to capture characteristics of ultrasonically activated soft tissue deformation and temperature evolution. At the cellular level, evaporation of water below the boiling temperature under ambient conditions is indicative of protein denaturation around the temperature threshold for coagulation of tissues. Further, with increasing operating frequency (or loading), the temperature rises faster leading to rapid evaporation of tissue cavity water, which may lead to accelerated protein denaturation and coagulation.

Keywords

multi-physics modeling; cavitation effect; equation-of-state; ultrasonically activated soft tissue; ultrasonically activated scalpel (UAS)

1. Introduction

Ultrasonic surgical instruments have been gaining popularity among surgeons in the last decade. An increasing number of surgical procedures including but not limited to head, neck, gynecological, colorectal and gastrointestinal surgeries are performed using ultrasonic surgical instruments (Sankaranarayanan et al., 2013). These instruments utilize ultrasonic vibrations to cut, coagulate and dissect tissues, and seal vessels. They have been proven to

*Corresponding author, Tel: +1 (518) 276-6351; Fax: +1 (518) 276-6025; des@rpi.edu.

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be superior to conventional instruments and techniques such as electrosurgical and laser-based devices as they impose lesser thermal injury, desiccation and charring, lower mean blood loss during surgery, no risk of stray current, neuromuscular stimulation, lesser operation time and post-operative pain, and no smoke during the operation to occlude laparoscopic view (Amaral and Chrostek, 1997).

Despite increasing popularity of ultrasound-based surgical procedures, the effects of cellular level mechanisms on the thermomechanical response of ultrasonically activated soft tissues have not been adequately modeled. The purpose of this work is to develop a multi-physics computational model to study the effects of cavitation, due to large transient pressure changes, on the thermomechanical response of ultrasonically activated soft tissue.

An ultrasonic surgical device or ultrasonically activated scalpel (UAS) uses mechanical vibrations to cut and coagulate soft tissues. The mechanical energy of oscillation is transferred to the tissue in the form of heat due to structural, and viscoelastic heating, which increases the temperature of the tissue. The protein in soft tissue denatures at 353–373K, and when it cools down to 336K, a sticky coagulum forms, sealing off the coapted vessel. Furthermore, the vibration of the scalpel blade tip induces large transient pressure changes, which cause intracellular and cellular water to vaporize below the boiling temperature under ambient conditions (373K). This cavitation effect leads to rupture of cellular walls that result in very precise cutting and fragmentation of the tissue (Lee and Park, 1999), with much less thermal damage when compared to other surgical procedures such as electrocautery and laser ablation (Amaral and Chrostek, 1997), where tissue cavity water evaporates at 373K.

The technique has been proven to be effective in cutting and coagulating tissues (Amaral, 1994, 1995; Amaral and Chrostek, 1997). It is now been widely used in laparoscopic surgeries because of its advantages that are associated with its unique working modes. Most of the studies assert that the temperature of tissue in the application of UAS does not exceed 373K (Koch et al., 2003; Bandi et al., 2008). However, it may produce higher temperatures for high power levels and longer activation time (Emam and Cuschieri, 2003; Kim et al., 2008), rendering it inefficient in sealing medium-to-large-sized vessels (Hruby et al., 2007; Clements and Palepu, 2007; Harold et al., 2003; Wolfe et al., 1991).

Although clinical application of UAS has been well documented, the understanding of the effects of cellular level mechanisms on the thermomechanical response of the ultrasonically activated soft tissue is limited. Therefore, it is imperative to develop such understanding for predicting the clinical outcome of these procedures. In this paper, we propose a multi-physics model to study the effects of cellular level cavitation mechanism on the thermomechanical response of the ultrasonically excited soft tissue. A thermodynamically consistent continuum model for the soft tissue is presented within the framework of large deformation viscoelasticity. The cavitation model based equation-of-state (EOS) describes the thermodynamic states such as temperature, and pressure due to evaporation of water content in the tissue cavity by absorbing heat due to structural and viscoelastic heating. The thermodynamics of tissue cavity water-vapor system is modeled based on a micro-explosion-based physical model for the laser ablation process (Majaron et al., 1999), which was developed assuming a spherical cavity embedded in a non-absorbing elastic medium.

No such assumption has been made in our cavitation based model. Using the proposed model, the thermomechanical response of soft tissue is studied for the operational range of frequencies of harmonic oscillations and applied loads found in typical UAS.

This paper is organized as follows. The continuum scale thermomechanical model for the ultrasonically activated soft tissue along with the cavitation based EOS are presented in Section 2. The thermomechanical response of human skin tissue for the operational range of frequencies and applied load is studied in Section 3. Both, the uniaxial and lateral (simple shear) loading conditions are considered. Conclusions and the future scope of work follow in Section 4.

2. Continuum model for soft tissue

Let $\mathbf{x} = \varphi(\mathbf{X}, t) : \Omega_0 \times \mathbb{R}^+ \rightarrow \mathbb{R}^3$ be a nonlinear deformation map at time $t \in \mathbb{R}^+$ that maps point $\mathbf{X}$ in the reference configuration $\Omega_0 \in \mathbb{R}^d$ of the tissue at temperature T_0 to point $\mathbf{x}$ in the current configuration $\Omega \in \mathbb{R}^d$ at temperature T . The current configuration is described by the deformation gradient tensor $\mathbf{F} = \varphi(\mathbf{X}, t) / \mathbf{X}$ such that $\det \mathbf{F} > 0$. We introduce a kinematic split of $\mathbf{F}$ to define volumetric and deviatoric (volume preserving) response in the nonlinear regime as $\mathbf{F} = J^{1/3} \bar{\mathbf{F}}$, where $\bar{\mathbf{F}} = J^{-1/3} \mathbf{F}$, $J = \det \mathbf{F}$ and $\det \bar{\mathbf{F}} = 1$. Associated with the right Cauchy-Green deformation tensor $\mathbf{C} = \mathbf{F}^T \mathbf{F}$ and $\bar{\mathbf{C}} = J^{-2/3} \mathbf{C}$ we define the Green-Lagrange strain measure as $\mathbf{E} = \frac{1}{2} (\mathbf{C} - \mathbf{I})$ and $\bar{\mathbf{E}} = \frac{1}{2} (\bar{\mathbf{C}} - \mathbf{I})$, respectively. The work conjugate symmetric Piola-Kirchhoff stress measure is defined as $\mathbf{S} = J \mathbf{F}^{-1} \boldsymbol{\sigma} \mathbf{F}^{-T}$, where $\boldsymbol{\sigma}$ is the Cauchy stress tensor.

With regard to the setup of the initial boundary value problem, we consider the field equations of elasticity (Gurtin, 1981) which will be solved for the oscillating displacement boundary condition and applied load using viscoelastic material model outlined in Section 1.1. The temperature evolution equation is given in Section 1.2. The cavitation based equation-of-state is derived in Section 1.3.

1.1. Viscoelastic material model

For the ultrasonically activated soft tissue, we postulate an uncoupled form of the free energy functional as

$$\psi(\mathbf{E}, \mathbf{Q}, T) = \psi^0(J, T) + \bar{\psi}^0(\bar{\mathbf{E}}, T) - \mathbf{Q} : \bar{\mathbf{E}} + \psi_I(\mathbf{Q}, T), \quad (1)$$

where $\mathbf{Q}$ is an internal variable and $\psi_I(\mathbf{Q}, T)$ is given by the Legendre transformation

$$\psi_I(\mathbf{Q}, T) = - (1 - \gamma) \bar{\psi}^0(\bar{\mathbf{E}}, T) + \bar{\mathbf{E}} : \mathbf{Q}, \text{ where } \gamma \in [0, 1) \text{ is a given parameter (Simo,}$$

1987). The Clausius-Duhem inequality $\dot{\psi} + \eta \dot{T} - \frac{1}{2} \mathbf{S} : \dot{\mathbf{C}} \leq 0$ (Coleman and Gurtin, 1967), where η denotes specific entropy per unit referential volume, leads to

$$\mathbf{S} = \frac{\partial \psi(\mathbf{E}, \mathbf{Q}, T)}{\partial \mathbf{E}}, \quad \eta = - \frac{\partial \psi(\mathbf{E}, \mathbf{Q}, T)}{\partial \theta}, \quad \text{and} \quad - \frac{\partial \psi(\mathbf{E}, \mathbf{Q}, T)}{\partial \mathbf{Q}} : \dot{\mathbf{Q}} \geq 0. \quad (2)$$

From Eq. (1) and (2¹) we obtain the following constitutive equation for the second Piola-Kirchhoff stress tensor (Lubliner, 2008)

$$\mathbf{S} = Jp\mathbf{C}^{-1} + J^{-2/3} \text{Dev} \left[\frac{\partial \bar{\psi}^0(\bar{\mathbf{E}}, T)}{\partial \bar{\mathbf{E}}} - \mathbf{Q} \right], \quad (3)$$

where $p = \Psi^0(J, T)/J$ denotes the hydrostatic pressure and 'Dev' is the Lagrangian deviator operator with $\text{Dev}[\cdot] = (\cdot) - \frac{1}{3}[\mathbf{C} : (\cdot)]\mathbf{C}^{-1}$.

The viscoelastic behavior is introduced using a linear rate equation for the internal variable as (Simo, 1987)

$$\dot{\mathbf{Q}} + \frac{1}{\nu} \mathbf{Q} = \frac{(1-\gamma)}{\nu} \text{Dev} \left[\frac{\partial \bar{\psi}^0(\bar{\mathbf{E}}, T)}{\partial \bar{\mathbf{E}}} \right], \quad \text{with } \mathbf{Q}(t=0) = \mathbf{0}, \quad (4)$$

where $\nu \in [0, \alpha]$ is the relaxation time. Substitution of Eq. (4) into (3) then yields the following equivalent expressions in terms of the symmetric Piola-Kirchhoff tensors,

$$\mathbf{S} = Jp\mathbf{C}^{-1} + J^{-2/3} \text{Dev}[\mathbf{H}], \quad \mathbf{H} := \int_0^t \left\{ \gamma + (1-\gamma)e^{-(t-s)/\nu} \right\} \frac{d}{ds} \left\{ \text{Dev} \left[\frac{\partial \bar{\psi}^0(\bar{\mathbf{E}}(s), T)}{\partial \bar{\mathbf{E}}} \right] \right\} ds \quad (5)$$

1.2. Temperature Evolution

In the absence of heat transfer or internal energy source in the system, the temperature evolution is a combination of purely structural and viscoelastic heating of the tissue. The temperature evolution follows from the energy balance as (Holzapfel and Simo, 1996)

$$\rho c_F \dot{T} = - \frac{1}{2} \rho c_F T \mathbf{\Gamma} : \dot{\mathbf{C}} + \frac{1}{\tilde{\eta}} |\mathbf{Q}|^2; \quad |\mathbf{Q}| = (\mathbf{Q} : \mathbf{Q})^{1/2}, \quad (6)$$

with the relationship $\tilde{\eta} = 2\mu\nu$ and $\mathbf{\Gamma} = : \beta / \rho c_P$ (Wallace, 1972), where c_F denote specific heat capacity at constant deformation, $\tilde{\eta}$ characterizes the viscosity of the relaxation process in terms of temperature dependent shear modulus μ and relaxation time $\nu = 0.25/f$, where f is the frequency of oscillation, and $\mathbf{\Gamma}$ denotes the Grüneisen tensor which is defined in terms of the fourth order elasticity tensor, coefficient of thermal expansion β , specific heat capacity at constant pressure c_P and density of the tissue ρ . The first and second terms in the Eq. (6) contribute to structural and viscoelastic heating, respectively.

1.3. Equation-of-state

On a more microscopic level, ultrasonically activated soft tissue produces large transient pressure changes which cause intracellular and cellular water to vaporize below the boiling temperature under ambient conditions. The cavitation effects lead to rupture of the cellular walls, resulting in precise cutting (Lee and Park, 1999). As a simplest starting point, we postulate that the tissue water is trapped in microscopic cavities in a non-absorbing elastic medium. The cavity water-vapor system is assumed to be associated with the continuum scale material point $\mathbf{X} \in \Omega_0$ of the tissue. As shown in Figure 1, initially, the cavity of volume V_0 is filled with water at body temperature (310K) and atmospheric pressure (101,325Pa). As the water absorbs the heat generated due to structural and viscoelastic heating (Section 1.2) of the ultrasonically excited soft tissue, its temperature increases until the boiling point T_0 is reached (corresponding to the cavity pressure p_0 as computed from Eq. (3)). Further heating converts a fraction (x) of the water mass into vapor and causes the cavity to expand under added pressure. The current volume V of the cavity can be calculated by adding volume of the water (V_w) and vapor (V_v) as

$$V = V_w + V_v = V_0 \left[\left(\frac{rT\rho_w}{p} - 1 \right) x + 1 \right], \quad (7)$$

where r is the specific gas constant for the water vapor, ρ_w denotes the density of liquid water at current pressure p and temperature T . Here, we assume that water vapor follows the ideal gas equation-of-state. To derive Eq. (7), we have used mass conservation $\rho_w V_0 = \rho_w V_w + \rho_v V_v$, where ρ_v is the density of water vapor, related to the cavity pressure (p) through the ideal gas law $p = \rho_v r T$ and vapor content (i.e. mass fraction of water converted to vapor) $x = \rho_v V_v / \rho_w V_0$. Notice that this change in the cavity volume is related to the Jacobian of deformation gradient tensor at the associated material point of the tissue as $J \equiv V / V_0$.

Further, an analytical relation can be obtained from the Clausius-Clapeyron equation for the ideal gas approximation of vapor much below the critical temperature that relates the equilibrium vapor pressure p and temperature T to the temperature-dependent latent heat of evaporation $q(T)$ as (Potter and Somerton, 1993)

$$\frac{dp}{dT} = \frac{pq(T)}{rT^2}. \quad (8)$$

For the average value of $q(T)$ on the interval $[T_0, T]$, denoted as $\bar{q}$, the integration of Eq. (8) results in an explicit relation between equilibrium temperature and pressure

$$\frac{1}{T} = \frac{1}{T_0} - \frac{r}{\bar{q}} \ln \left(\frac{p}{p_0} \right). \quad (9)$$

Eqs. (7), and (9) will lead to an equation-of-state (EOS) that relates cavity pressure p with the relative vapor content x as

$$x = \left(\frac{V}{V_0} - 1 \right) \left[\frac{rT_0\rho_w}{p} \left\{ 1 - \frac{rT_0}{\bar{q}} \ln \left(\frac{p}{p_0} \right) \right\}^{-1} - 1 \right]^{-1}. \quad (10)$$

This EOS together with the following energy balance equation for the cavity-water-vapor system

$$\Delta Q = \rho [wq(T) \Delta x + wx(c_v + r) \Delta T + w(1-x)c_w \Delta T + (1-w)c_s \Delta T] \quad (11)$$

describes the evolution of local thermodynamic variables at each macroscopic material point $\mathbf{X} \in \Omega_0$, where Q is the total heat addition to the system that includes structural and viscoelastic heating contributions as discussed in Section 1.2, w is the water content, and c_s , c_w , and c_v denotes specific heat capacity of solid, water, and vapor part, respectively. In Eq (11), the first three terms describe the enthalpy changes in additional evaporation, and heating of vapor and water to the new equilibrium temperature, respectively. The last term accounts for eventual conductive thermalization of the solid tissue components.

The energy balance equation (Eq. (11)) combined with the EOS (Eq. (10)) gives a highly nonlinear set of equations, which is solved for the equilibrium temperature T using Newton's method.

3. Numerical examples

We investigate the effects of cellular level cavitation mechanism on the thermomechanical response of human skin using a continuum model for soft tissue and cavitation based equation-of-state (EOS) presented in previous Section 2. The model parameters including thermal and material properties of human skin are summarized in Table 1. The elastic constitutive equation is furnished by the following stored free energy functional

$$\psi^0(\mathbf{C}, T) = \frac{k}{2}(J-1)^2 + \frac{\mu}{2}(\text{tr} \bar{\mathbf{C}} - 3), \quad (12)$$

where μ and κ are material properties, and $p = k(J-1)$ defines the hydrostatic pressure.

A finite element computational model has been developed to simulate the action of (a) ultrasonic knife on the tissue plane normal to loading direction (Figure 2a), and (b) ultrasonically activated scalpel (UAS) on a tissue block (Figure 6). In the absence of frictional sliding between tissue and the ultrasonic device, this is equivalent to uniaxial, and lateral (simple shear) oscillation of the computational model of the tissue, respectively. The initial temperature of the tissue is assumed to be the body temperature (310K), whereas, tissue cavities are initially filled with water at body temperature and atmospheric pressure (101,325Pa). A sinusoidal oscillation $u = u_0 \sin(\omega t)$ is applied to the tissue, where u_0 and $\omega = 2\pi f$ denote the amplitude and angular frequency of oscillation, respectively. For the ultrasonically activated devices, the amplitude of oscillation ranges from 60–100 μm . Unless

otherwise stated, the ultrasonic device is assumed to vibrate at natural frequency of 55.5kHz and 80 μ m amplitude.

A large deformation analysis using a total Lagrangian approach (Bathe, 1995) is performed for the solution of non-linear initial boundary value problem associated with the field equations of elasticity. The macroscopic domain of the soft tissue is discretized using tri-linear hexahedral finite elements with eight Gauss points per element. The thermodynamic states (p , T) at the macroscale integration point are obtained by solving the energy balance equation (Eq. (11)) along with the EOS (Eq. (10)) numerically using Newton's method. The transient response of the ultrasonically vibrating soft tissue is obtained by solving continuum momentum balance equation using an explicit central difference time integration scheme, while inertial effects are eliminated by mass scaling.

In the following Sections 3.1 and 3.2, we present representative example problems that simulate thermomechanical behavior of soft tissues subject to uniaxial and lateral (simple shear) ultrasonic vibrations.

3.1. Uniaxial oscillation

The thermomechanical response of the action of ultrasonic knife on the tissue plane normal to the loading direction is studied using a single tri-linear hexahedral finite element with eight Gauss points. A uniaxial sinusoidal deformation $u_2 = u_0 \sin(\omega t)$ is applied normal to one of the faces of a unit *millimeter* cube such that the face initially normal to the loading axis remains normal to it (Figure 2a). Plots are obtained at an integration point with the reference coordinate (0.789mm, 0.789mm, 0.211mm) of the unit cube. The purpose of this study is to examine the efficacy of the EOS (Eq. (10)) in capturing cavitation effect in ultrasonically activated soft tissue. A representative example of the action of UAS on the soft tissue is presented in the next section.

We begin by studying the response of tissue subjected to uniaxial oscillatory displacement at 55.5kHz frequency and 80 μ m amplitude. The deformation history and temperature distribution at various time steps are plotted in Figure 2b–e. Here, we clearly see that the action of the vibrating knife not only increases the temperature of the tissue but also inflates it. To understand this phenomenon, in Figures 3a–d, we plot the evolution of von Mises stress, pressure, temperature, and vapor content with time. While the pressure and temperature plots in Figures 3b and 3c, follow from the Clausius-Clapeyron equation (Eq. (9)), the evaporation and condensation cycles in the vapor content curve clearly show the effects of oscillating volumetric changes (Eq. (10)) and stresses (Figure 3a). From Figure 3d, we see that water starts to evaporate at body temperature (Figure 3c), when the cavity pressure is much below the atmospheric pressure (Figure 3b). Further, from Figure 3c, we observe that the peak temperature for the simulation time is within the range of temperatures required for protein denaturation (353–373°K) leading to tissue coagulation. The deformation history in Figure 2b–e indicates volumetric expansion of the block due to increasing pressure caused by cavitation induced evaporation of water. This may lead to fragmentation of tissue due to rupture of cellular walls under exponentially increasing cavity pressure.

Next, we study the effect of oscillation frequencies of the ultrasonic device on the pressure and temperature evolution in the tissue block and the vaporization of water in the cavity with time. Five different equally spaced frequencies ranging from 35kHz to 75kHz are considered. The amplitude of vibration is kept constant at 80 μ m. The cumulative non-negative viscoelastic dissipation due to the non-equilibrium stresses (Eq. (6)) increases with increasing frequency of oscillation leading to more viscoelastic heating, resulting in faster temperature rise at higher frequencies (Figure 4b). On the contrary, purely structural heating has no net heating effect over a complete oscillation cycle. Further, faster temperature rise at sub-atmospheric cavity pressure leads to faster evaporation of cavity water (Figure 4c) resulting in rapid increase in cavity pressure (Figure 4a).

The effect of amplitude of oscillation on the evolution of pressure and temperature in the tissue, and evaporation of cavity water is investigated at 55.5kHz frequency. Five different equally spaced operational amplitudes ranging from 60 μ m to 100 μ m are considered. With increasing amplitude of oscillation, viscoelasticity leads to enhanced structural heating, as well as larger viscoelastic heating due to accumulated viscoelastic dissipation, see Eq. (6). This results in faster temperature rise with increasing amplitude of vibration (Figure 5b) leading to rapid evaporation of cavity water (Figure 5c) at a pressure below the atmospheric pressure. The faster evaporation of cavity water results in rapid increase in cavity pressure (Figure 5a). While faster temperature rise may initiate early protein denaturation and coagulation of tissue, the rapid cavity pressure buildup may have significant effect on tissue cutting time.

3.2. Lateral oscillation

A computational model of a rectangular tissue block with dimension 6.0 $\times$ 2.0 $\times$ 1.0mm³ is discretized using 12,000 tri-linear hexahedral finite elements with eight Gauss points per element. A lateral sinusoidal displacement $u_1 = u_0 \sin(\omega t)$ is applied in the middle of the top surface, while the corresponding bottom surface is constrained in the x_2 -direction (Figure 6). This type of boundary condition simulates the jaws of the scalpel that clutches the tissue and prevents it from moving. A pressure load is applied to the middle of the top surface of the tissue to simulate the applied pressure load (Figure 6). Plots are obtained at an integration point with the reference coordinate (0.021mm, 0.979mm, 3.079mm) of the tissue block as shown in Figure 6. The finite element with this integration point falls within the region where transient pressure changes are maximum leading to strong cavitation effect as we will see later in this section.

First, we study the thermomechanical response of the tissue subjected to lateral oscillation at 55.5kHz frequency and 80 μ m amplitude, and 2.0atm applied pressure load. Figures 7a–d shows the evolution of von Mises stress, pressure, temperature, and vapor content with time. As in the uniaxial loading case, the cavity water starts evaporating at body temperature (Figure 7d) and peak temperature in the simulated time is within the range of protein denaturation and coagulation temperature. The evaporation and condensation cycles in the vapor content-time trace in Figure 7d corresponds to the oscillating cavity pressure in Figure 7b, i.e., when the cavity pressure is above the atmospheric pressure and the temperature is below the boiling temperature under ambient conditions (Figure 7c), a fraction of the vapor

condenses back to water (Figure 7d). On the other hand, when the cavity pressure drops below the atmospheric pressure (Figure 7b) and there is sufficient thermomechanical heat to overcome the enthalpy of evaporation based on energy balance equation (Eq. (11)), water evaporates (Figure 7d). This indicates that the running minimum¹ cavity pressure and temperature dictates the maximum vapor content within each cycle. From Figures 7b–d, we see that as the running minimum pressure and equilibrium temperature gradually increase with time, the running maximum¹ vapor content also increases. Therefore, to study the effects of vibrational frequency and applied load on the thermomechanical response of tissue, we compare running minimum pressure and running maximum vapor content next.

The effect of applied pressure loads on the evolution of pressure and temperature in the tissue body, and evaporation of cavity water is investigated at 55.5kHz frequency and 80 μ m amplitude of oscillation. Four different pressure loads ranging from 1.0atm to 4.0atm, in increments of 1.0 atm, are considered. In Figures 8a and 8c, running minimum pressure and maximum vapor content within each cycle is compared for these four applied pressure loads, respectively. With increasing applied load, viscoelastic deformation of the tissue results in more structural heating within a cycle, as well as larger viscoelastic heating due to accumulated non-negative viscoelastic dissipation as described by Eq. (6). This results in faster temperature rise (Figure 8b) leading to rapid evaporation of cavity water (Figure 8c) for the running minimum pressure below atmospheric pressure. While faster temperature rise may initiate early protein denaturation and coagulation, the rapid cavity pressure buildup may significantly effect tissue cutting time.

Next, we study the effect of oscillation frequencies of the ultrasonic device on the pressure and temperature evolution in the tissue block and the vaporization of water in the cavity. Four different frequencies ranging from 35kHz to 95kHz, in increments of 20 kHz, are considered. The amplitude of vibration is kept constant at 80 μ m. A constant pressure load of 2.0 atm is applied to the tissue. The cumulative viscoelastic dissipation increases with increasing frequency of oscillation leading to more viscoelastic heating, resulting in steeper temperature rise at the higher frequencies (Figure 9b). Further, faster temperature rise at sub atmospheric running minimum cavity pressure leads to rapid increase in the running maximum vapor content (Figure 9c), resulting in rapid increase in cavity volume as defined by Eq. (7). These trends are very similar to that of the uniaxial loading case (Figure 4).

The distribution of the von Mises stress, pressure, temperature, and vapor content are presented in Figure 10. Contour plots are taken at 1.25ms for the tissue activated at 55.5kHz frequency and 80 μ m amplitude of oscillation, and for the pressure load of 2.0atm. From the vapor content distribution in Figure 10d, we observe that more cavity water evaporates at temperatures below the boiling point under ambient conditions along the side that is under tensile shear (Figure 10a). On the compressive shear side cavity water evaporates above ambient boiling temperature (Figure 10c and 10c). Higher evaporation of cavity water results in increase in cavity pressure along the tensile shear side (Figure 10b).

¹'Running maximum' and 'running minimum' of a physical quantity are defined as the maximum and minimum values of the time evolving oscillatory physical quantity within each cycle.

4. Conclusions

A multi-physics model that accounts for the cellular level cavitation effects is proposed to describe the thermomechanical response of ultrasonically activated soft tissue. A thermodynamically consistent continuum model for the soft tissue is presented within the framework of large deformation viscoelasticity. The cavitation model based equation-of-state describes the cavity temperature, and pressure due to evaporation of cavity water by absorbing heat due to structural and viscoelastic heating. The evolution of thermodynamic states such as pressure and temperature, and characteristics of vaporization of water (vapor content) of the tissue are studied for representative operational range of oscillation frequencies and loading conditions.

The model is shown to successfully capture the characteristics of ultrasonically activated soft tissue deformation and temperature evolution as well as vaporization of water below ambient boiling temperature. The contour plots show localization of temperature and vapor content. We also observe that the higher the frequency of oscillation of the ultrasonic device, the faster the temperature rise in the tissue and the earlier the water starts to vaporize. Another interesting observation is that higher loading, which corresponds to greater grasping force on the tool, leads to faster temperature rise and accelerated water vaporization. Early evaporation of water may be hypothesized to lead to accelerated protein denaturation and coagulation leading to precise cutting and fragmentation of tissue. This observation remains to be validated through experimentation.

The proposed model can be further extended to include frictional heating due to interaction of tissue with the ultrasonic devices. For a more complete description of the tissue degradation due to cellular wall rupture, a damage model needs to be employed. A coupled thermal and mechanical analysis would allow studying the response of UAS for longer time interval, where deformation process is no longer adiabatic and thermal conduction needs to be incorporated. However, excessive distortion of elements due to sharp increase in cavity (hydrostatic) pressure is likely to occur for longer simulation times. Further, the cavitation based EOS model presented in this work assumes equilibrium of water with the moist environment. It does not consider transport of water-vapor mixture, which could have significant effect on the overall response of tissue subjected to ultrasonic vibrations. Hence, a more comprehensive model needs to be developed that captures these effects associated with ultrasonically activated soft tissue.

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References

- Amaral JF. The experimental development of an ultrasonically activated scalpel for laparoscopic use. *Surgical Laparoscopy Endoscopy & Percutaneous Techniques*. 1994; 4(2):92–99.
- Amaral JF. Laparoscopic cholecystectomy in 200 consecutive patients using an ultrasonically activated scalpel. *Surgical Laparoscopy Endoscopy & Percutaneous Techniques*. 1995; 5(4):255–262.

- Amaral JF, Chrostek CA. Experimental comparison of the ultrasonically-activated scalpel to electrosurgery and laser surgery for laparoscopic use. *Minimally Invasive Therapy & Allied Technologies*. 1997; 6(4):324–331.
- Bandi G, Wen CC, Wilkinson EA, Hedican SP, Moon TD, Nakada SY. Comparison of blade temperature dynamics after activation of Harmonic Ace scalpel and the Ultracision Harmonic Scalpel LCS-K5. *Journal of Endourology*. 2008; 22(2):333–336. [PubMed: 18294041]
- Bathe, J. *Finite Element Procedures*. Prentice-Hall: Englewood Cliffs; New Jersey: 1995.
- Clements RH, Palepu R. In vivo comparison of the coagulation capability of SonoSurg and Harmonic Ace on 4 mm and 5 mm arteries. *Surgical Endoscopy*. 2007; 21(12):2203–2206. [PubMed: 17479325]
- Coleman BD, Gurtin ME. Thermodynamics with internal state variables. *The Journal of Chemical Physics*. 1967; 47(2):597–613.
- Duck, FA. *Physical Properties of Tissue*. Academic Press; London: 1990.
- Emam TA, Cuschieri A. How safe is high-power ultrasonic dissection? *Annals of Surgery*. 2003; 237(2):186–191. [PubMed: 12560776]
- Gurtin, ME. *An introduction to continuum mechanics*. Academic Press; New York: 1981.
- Harold KL, Pollinger H, Matthews BD, Kercher KW, Sing RF, Heniford BT. Comparison of ultrasonic energy, bipolar thermal energy, and vascular clips for the hemostasis of small-, medium-, and large-sized arteries. *Surgical Endoscopy and Other Interventional Techniques*. 2003; 17(8):1228–1230. [PubMed: 12799888]
- Holzapfel GA, Simo JC. A new viscoelastic constitutive model for continuous media at finite thermomechanical changes. *International Journal of Solids and Structures*. 1996; 33(20):3019–3034.
- Hruby GW, Marruffo FC, Durak E, Collins SM, Pierorazio P, Humphrey PA, Mansukhani MM, Landman J. Evaluation of surgical energy devices for vessel sealing and peripheral energy spread in a porcine model. *The Journal of Urology*. 2007; 178(6):2689–2693. [PubMed: 17945280]
- Kim FJ, Chammas MF Jr, Gewehr E, Morihisa M, Caldas F, Hayacibara E, Baptistussi M, Meyer F, Martins AC. Temperature safety profile of laparoscopic devices: Harmonic ACE (ACE), Ligasure V (LV), and plasma trisector (PT). *Surgical Endoscopy*. 2008; 22(6):1464–1469. [PubMed: 18027042]
- Koch C, Friedrich T, Metternich F, Tannapfel A, Reimann HP, Eichfeld U. Determination of temperature elevation in tissue during the application of the harmonic scalpel. *Ultrasound in Medicine & Biology*. 2003; 29(2):301–309. [PubMed: 12659918]
- Lee SJ, Park KH. Ultrasonic energy in endoscopic surgery. *Yonsei Medical Journal*. 1999; 40(6):545–549. [PubMed: 10661030]
- Lemmon, EW.; McLinden, MO.; Friend, DG. *NIST Chemistry Webbook, NIST Standard Reference Database*. Vol. 69. National Institute of Standards and Technology; Gaithersburg MD: 2005. Thermophysical properties of fluid systems.
- Lipkin M, Hardy JD. Measurement of some thermal properties of human tissues. *Journal of Applied Physiology*. 1954; 7(2):212–217. [PubMed: 13211500]
- Lubliner, J. *Plasticity theory*. Dover Publications, Mineola; New York: 2008.
- Majaron B, Plestenjak P, Luka M. Thermo-mechanical laser ablation of soft biological tissue: modeling the micro-explosions. *Applied Physics B*. 1999; 69(1):71–80.
- Potts RO, Chrisman DA, Buras EM. The dynamic mechanical properties of human skin in vivo. *Journal of Biomechanics*. 1983; 16(6):365–372. [PubMed: 6619155]
- Potter, MC.; Somerton, CW. *Theory and Problems of Engineering Thermodynamics*. MacGraw-Hill; New York: 1993.
- Sankaranarayanan G, Resapu RR, Jones DB, Schwaitzberg S, De S. Common uses and cited complications of energy in surgery. *Surgical Endoscopy*. 2013; 27(9):3056–3072. [PubMed: 23609857]
- Simo JC. On a fully three-dimensional finite-strain viscoelastic damage model: formulation and computational aspects. *Computer Methods in Applied Mechanics and Engineering*. 1987; 60(2):153–173.

- Wallace, DC. Thermodynamics of Crystals. John Wiley & Sons; New York: 1972.
- Wolfe BM, Gardiner BN, Leary BF, Frey CF. Endoscopic cholecystectomy: an analysis of complications. Archives of Surgery. 1991; 126(10):1192–1198. [PubMed: 1834039]
- Xu F, Lu TJ, Seffen KA. Biothermomechanics of skin tissues. Journal of the Mechanics and Physics of Solids. 2008; 56(5):1852–1884.

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Highlights

1. Multi-physics modeling of ultrasonically activated soft tissue
2. Model accounts for the cellular level cavitation effects to describe the thermomechanical response of soft tissue subject to ultrasonic oscillations
3. Model is shown to capture the characteristics of ultrasonically activated soft tissue deformation, temperature evolution, and vaporization of water below 373K
4. Evaporation of water below 373K is indicative of protein denaturation around the temperature threshold for coagulation of tissues
5. With increasing operating frequency (or loading), the temperature rises faster leading to rapid evaporation of tissue cavity water, which may lead to accelerated protein denaturation and coagulation

Figure 1. Schematic showing cavitation induced evaporation of water and changes in thermodynamic states, when mechanical heat is added to the cavity-water-vapor system.

Figure 2.

(a) Schematic of uniaxial oscillation of tissue. Temperature distribution and deformation history shown at (b) 1.2ms, (c) 1.5ms, (c) 1.7ms, and (d) 1.8ms.

Figure 3.

Evolution of (a) von Mises stress, (b) pressure, (c) temperature, and (d) vapor content with time. Plots are obtained at the centroid of the unit cube for the ultrasonic knife operating at 55.5kHz frequency and 80 μ m amplitude.

Figure 4. Evolution of (a) pressure, (b) temperature, and (c) vapor content with time for the tissue operated at various ultrasonic frequencies and 80 μ m amplitude of oscillation.

Figure 5.

Evolution of (a) pressure, (b) temperature, and (c) vapor content with time for the tissue operated at 55.5kHz ultrasonic frequency and various amplitudes of oscillations.

Figure 6.
Computational model and boundary conditions.

Figure 7.

Time evolution of (a) von Mises stress, (b) pressure, (c) temperature, and (d) vapor content for the UAS operating on the tissue block at 55.5kHz frequency and 80 μ m amplitude subjected to 2.0atm pressure load.

Figure 8.

Time evolution of (a) running minimum pressure, (b) temperature, and (c) running maximum vapor content for the tissue operated at 55.5kHz ultrasonic frequency and 80 μ m amplitude, and various applied pressure loads.

Figure 9.

Evolution of (a) running minimum pressure, (b) temperature, and (c) running maximum vapor content with time for the tissue operated at various ultrasonic frequencies and 80 μ m amplitude of oscillation, and subjected to 2.0 atm pressure load.

Figure 10.

Distribution of (a) von Mises stress, (b) pressure, (c) temperature, and (d) vapor content at 1.25ms. Contour plots are shown for the tissue activated at 55.5kHz frequency and 80 μ m amplitude of oscillation, and 2.0atm applied pressure load.

Table 1

Thermal and mechanical properties of human skin.

Young's modulus ¹ (E)	0.5 MPa
Poisson's ratio ² (ν)	0.48
Average density ³ (ρ)	$1.1 \times 10^3 \text{ kg.m}^{-3}$
Coefficient of thermal expansion ² (β)	$1.0 \times 10^{-4} \text{ K}^{-1}$
Specific heat capacity of tissue ⁴ (c_p)	$3.39 \times 10^3 \text{ J.kg}^{-1}.\text{K}^{-1}$
Water content ¹ (w)	70%
Liquid water density ⁵ (ρ_w)	$1.0 \times 10^3 \text{ kg.m}^{-3}$
Specific heat capacity of solid part ¹ (c_s)	$1.7 \times 10^3 \text{ J.kg}^{-1}.\text{K}^{-1}$
Specific heat capacity of water part ⁵ (c_w)	$4.187 \times 10^3 \text{ J.kg}^{-1}.\text{K}^{-1}$
Specific heat capacity of vapor part ⁵ (c_v)	$1.996 \times 10^3 \text{ J.kg}^{-1}.\text{K}^{-1}$
Specific gas constant for water vapor ⁵ (r)	$4.620 \times 10^2 \text{ J.kg}^{-1}.\text{K}^{-1}$
Latent heat of vaporization of water ⁵ ($\bar{q}$)	$2.383 \times 10^6 \text{ J.kg}^{-1}$
Viscoelastic parameter (γ)	0.3

¹Duck (1990);²Xu et al. (2008);³Potts et al. (1983);⁴Lipkin and Hardy (1954);⁵Lemmon et al. (2005)