On the Thermal Effects Associated with Radiation Force Imaging of Soft Tissue
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Abstract—Several laboratories are investigating the use of acoustic radiation force to image the mechanical properties of tissue. Acoustic Radiation Force Impulse (ARFI) imaging is one approach that uses brief, high-intensity, focused ultrasound pulses to generate radiation force in tissue. This radiation force generates tissue displacements that are tracked using conventional correlation-based ultrasound methods. The tissue response provides a mechanism to discern mechanical properties of the tissue.

The acoustic energy that is absorbed by tissue generates radiation force and tissue heating. A finite element methods model of acoustic heating has been developed that models the thermal response of different tissues during short duration radiation force application. The beam sequences and focal configurations used during ARFI imaging are modeled herein; the results of these thermal models can be extended to the heating due to absorption associated with other radiation force-based imaging modalities. ARFI-induced thermal diffusivity patterns are functions of the transducer f-number, the tissue absorption, and the temporal and spatial spacing of adjacent ARFI interrogations. Cooling time constants are on the order of several seconds. Tissue displacement due to thermal expansion is negligible for ARFI imaging. Changes in sound speed due to temperature changes can be appreciable. These thermal models demonstrate that ARFI imaging of soft tissue is safe, although thermal response must be monitored when ARFI beam sequences are being developed.

I. INTRODUCTION

The use of acoustic radiation force to interrogate the mechanical properties of soft tissues is becoming a widely investigated research area. In general, acoustic radiation force is used to excite tissue, and the tissue response is monitored using either ultrasonic or magnetic resonance methods. For an analytic description of the mechanical response of soft tissue to focused acoustic radiation force, the reader is referred to Sarvazyan et al. [2].

There are many methods using acoustic radiation force currently under investigation. Vibroacoustography uses frequency-shifted, confocal beams to generate an oscillating radiation force within tissues, and the tissue response is monitored either with a hydrophone [3], [4], or by ultrasonic methods [5]. The KAVE method uses radiation force to generate a steady-state stress within soft gels and the vitreous humor of the eye, and ultrasonic displacements are tracked using an ultrasound detector. This vibration is driven by the rapid oscillations of the surrounding medium, which are caused by the propagation of elastic waves generated by the radiation force.

The purpose of the work herein is to determine the increase in tissue temperature that is associated with the pulse sequences used in single-location, two-dimensional, shear wave, and ARFI M-mode imaging in soft tissue. In addition, the impact of the heating on displacement tracking is evaluated with respect to thermal expansion and sound speed changes. Nomenclature and physical constants used in this paper are given in Table I.

II. BACKGROUND

A. ARFI Imaging

ARFI imaging is a radiation force-based imaging method that studies the local mechanical properties of...
tissue [16]. ARFI imaging uses short-duration (<1 ms), high-intensity acoustic pulses to generate localized displacements in tissue, and the tissue recovery response is monitored using ultrasonic correlation-based methods [17]. Images of two-dimensional ROIs are generated by sequentially interrogating multiple lateral locations, as is done in color Doppler imaging. During the application of high-intensity pulses, energy is absorbed by the tissue that results in the generation of acoustic radiation force, tissue displacement, and tissue heating.

Herman et al. [18] recently published a study investigating ultrasonic imaging modalities that use short-duration, high-intensity pulses, such as streaming detection [19] and shear wave ARFI imaging [11]. This study demonstrates, for moderately absorbing soft tissues (0.3–0.5 dB/cm/MHz), that repeated interrogations in a given location result in peak heating at the focus versus the tissue surface. Short duration, in situ intensities higher than the currently accepted steady-state limit of 0.72 W/cm² are shown to be reasonable in soft tissue, but the presence of bone interfaces increases the potential for thermal tissue damage.

B. Bio-Heat Transfer Equation

As an acoustic wave propagates through a dissipative medium, an energy gradient is established in the medium, arising from either absorption or reflection of the wave. This energy gradient applies a force in the direction of wave propagation, and the absorption of energy results in the generation of heat in the tissue. The temperature rise in tissue can be estimated with the linear bio-heat transfer equation [1], [20], [21]:

\[ \dot{T} = \kappa \nabla^2 T - \frac{T}{\tau} + \frac{q_v}{c_v}, \]  

(1)

where \( T \) [°C] is the temperature, \( \dot{T} \) [°C/s] is the time rate change in temperature, \( \kappa \) [0.00143 cm²/s] is the thermal diffusivity for soft tissue, \( \tau \) [s] is the perfusion time constant, \( q_v \) [J/cm³] is the rate of heat production per unit volume, and \( c_v \) [4,200 mW·s/cm³/°C] is the heat capacity per unit volume for soft tissue. Eq. (1) can be solved analytically [20] and/or numerically [1], [18], [22]–[24], and it has been validated experimentally by several groups [1], [23], [24]. A summary of much of the work done on acoustic heating is provided in [1]. For ARFI imaging, the duration of heat application is very short (<1 ms), thus the effects of perfusion can be neglected [1], [20], simplifying (1) to:

\[ \dot{T} = \kappa \nabla^2 T + \frac{q_v}{c_v}. \]  

(2)

The heat source function for a continuous wave field of ultrasound can be expressed as [1], [20], [22]:

\[ q_v = \frac{\alpha I_0}{\rho c}, \]  

(3)

where \( \alpha \) [Np/cm] is the absorption coefficient for soft tissue, \( \rho \) [1.0 g/cm] is the density of soft tissue, \( c \) [1,540 m/s] is the speed of sound in soft tissue, and \( p_o \) [Pa] is the acoustic pressure. For a linearly traveling plane wave, (3) reduces to [1], [20]:

\[ q_v = 2\alpha I. \]  

(4)

where \( I \) [W/cm²] is the acoustic beam intensity.

For a focused beam, the distribution of the applied heat source \( q_v \) is dependent upon the focal configuration of the transducer, which is often characterized by the transducer f-number (F/#):

\[ F/# = \frac{z}{D}. \]  

(5)

where \( z \) [cm] is the acoustic focal length and \( D \) [cm] is the aperture width. This notation will be used throughout this paper.

C. Analytic Solutions to the Bio-Heat Transfer Equation

Analytic steady-state solutions to (1) exist for simple geometries. For example, the steady-state temperature rise for a heated sphere of radius \( R \) in an infinite medium is given by the expression [20]:

\[ \Delta T = \frac{\alpha I R^2}{K}, \]  

(6)

where \( K \) [6.0 mW/cm/°C] is the thermal conductivity of soft tissue. Eq. (6) is used to validate our simulation approach. In addition, the transient behavior of our model is validated using a simplified version of the bio-heat transfer equation (1), in which heat loss due to perfusion and thermal conduction is neglected. Eq. (1) becomes:

\[ \dot{T} = \frac{q_v}{c_v}. \]  

(7)

The solution to (7) provides a linear relationship between the increase in temperature, the application time, and the in situ intensity of the acoustic beam, as well as the absorption coefficient of the tissue:

\[ \Delta T = \frac{q_v t}{c_v} = \frac{2\alpha I t}{c_v}, \]  

(8)

where \( \Delta T \) [°C] is the increase in temperature and \( t \) [s] is the heat application time. Validation of the transient behavior of our thermal model is achieved by comparison with (8).

D. Thermal Expansion

Ultrasonic correlation-based methods are used to track radiation force-induced tissue displacements. Tissue displacement arising from thermal expansion can lead to speckle decorrelation and false estimates of displacement that could impact imaging performance. Linear thermal expansion in tissue can be characterized using the expression [25]:

\[ d(T) = d_o(1 + \beta(T - T_o)), \]  

(9)
where $T_o$ [°C] is the baseline temperature, $d_o$ [cm] is the baseline length of the region of interest, and $\beta [\frac{3}{1000} °C^{-1}]$ is the thermal expansion coefficient for soft tissue [1]. Eq. (9) is used to evaluate the effects of thermal expansion during ARFI imaging.

E. Temperature-Dependent Changes in Sound Speed

The sound speed of biological tissues is dependent on the temperature of the tissue [25], [26]. For temperature changes of less than 6°C, the change in sound speed with respect to temperature ($\frac{\partial c}{\partial T}$) is assumed to be linear, with typical values for $\frac{\partial c}{\partial T}$ being $-2.93$ m/s/°C for breast fat and $1.0$ m/s/°C for liver [25], [26]. False estimates of the sound speed due to heating can lead to over- or under-estimation of displacements in correlation-based methods [25], [26]. The finite element methods (FEM) model developed herein is used to evaluate the impact of sound speed changes on ARFI imaging.

III. Methods

A. Approach to Thermal Simulation

The FEM model of heating associated with ARFI imaging presented herein estimates increases in temperature resulting from acoustic energy absorption by tissue. Model implementation is performed using a two-step approach: first, the spatially distributed acoustic intensity field corresponding to the experimental transducer is simulated and the associated heat source function is computed using (4); second, finite element methods are used to determine the thermal response of the tissue.

B. Simulating Intensity Fields and Thermal Model Generation

FIELD II \(^1\) [27], a linear acoustic field simulation software program, is used to determine the acoustic intensity distribution associated with the high intensity pulses currently used for ARFI imaging (Siemens 75L40 linear transducer: one row of elements, excitation center frequency of 7.2 MHz, aperture width dependent on the number of active elements; Siemens Medical Systems, Ultrasound Group, Issaquah, WA). The transducer is simulated as F/3.8 in elevation and F/1.3 laterally, with a focal depth of 2.0 cm. This corresponds to the experimental setup currently utilized to perform ARFI imaging [28], and is the default transducer focal configuration, unless otherwise specified. An unfocused beam (i.e., 10 element transmit aperture fired simultaneously) is also simulated to compare and contrast the tissue heating associated with different focal configurations.

The three-dimensional intensity fields are computed and normalized, and a threshold of 5% of the maximum intensity is imposed to reduce computational overhead. This intensity field is converted to a heat source field using (4), imported into a finite element mesh generation program (HyperMesh, Altair Computing Inc., Troy, MI), and superimposed onto a rectangular solid mesh with a spatial extent of 7.5 mm (elevation) by 25 mm (lateral) by 37.5 mm (axial). These dimensions are chosen to be much larger than the heat source in order to allow for heat diffusion. Plane-symmetry in elevation is assumed in the model.

Fig. 1 illustrates the heat source function for ARFI imaging in tissue media with absorptions of 0.5 and 1.0 dB/cm/MHz for an F/1.3 focal configuration. The transducer is centered on the top, forward surface of the model and spans a total of 1.52 cm as delimited by the contours of the heat source function on the top surface of the right-hand figure. Since plane-symmetry is assumed, only half of the transducer is modeled in elevation. The front of the figure corresponds to an axial-lateral plane centered in elevation on the transducer face. This plane is the default plane on which temperature profiles are displayed, unless otherwise specified. Studies by Duck et al. and Herman et al. have shown that, for moderately attenuating tissues (0.5 dB/cm/MHz), repeated insonification at a given location results in peak heating at the focus, as opposed to the tissue surface [27], [29]. Heat generated at the face of the transducer, along with heat generated during the tracking of tissue displacement, is not included in these simulations.

The tissue is modeled as a thermally homogeneous, isotropic solid and is meshed with trilinear dynamic thermal elements. Thermal material properties, such as heat capacity [4,200 mW·s/cm\(^3\)/°C] and thermal diffusivity [0.00143 cm\(^2\)/s], are assumed to be independent of temperature, and therefore can be linearly scaled [20]. There are 220,313 nodes and 207,760 elements in the mesh, creating a mesh with cubes of equal volume and a uniform node spacing of 0.352 mm. All nodal temperatures in the model are initialized by default to 0°C (the environment temperature) as relative temperature changes are simulated with respect to baseline temperature. Normalized heat source values applied to elements from the simulated intensity outputs are scaled to a peak in situ intensity value of $I = 1,000$ W/cm\(^2\), corresponding to a peak heat source of 49.59 cal/s/cm\(^3\), when 69 elements are fired in an F/1.3 configuration. This value is consistent with in situ values indirectly measured during ARFI imaging [16] using a linear extrapolation of small-signal derated fields, which will overestimate the true focal fields [30]. A peak in situ intensity value of 1,000 W/cm\(^2\) is also implemented for the unfocused transducer configuration to generate displacements of at least 10 μm that are desirable for ARFI imaging [10]. The top model surface that represents contact with the environment/transducer is held at the environmental temperature, whereas the other boundaries of the model have insulating boundary conditions, simulating a continuum of tissue [22].

LS-DYNA3D (Livermore Software Technology Corporation, Livermore, CA) is used to solve for the dynamic temperature fields and the thermal expansion displace-

\(^1\)http://www.es.oersted.dtu.dk/staff/jaj/field/
Fig. 1. Heat source function associated with ARFI imaging for tissue absorptions of 0.5 and 1.0 dB/cm/MHz for an F/1.3 focal configuration. The heat source is shown relative to the spatial boundaries of the finite element mesh (7.5 mm by 25 mm by 37.5 mm). The transducer is located on the top, forward surface of the model, as delimited by the contours on the top surface of the figure on the right, for a total lateral dimension of 1.52 cm. Elevation symmetry is assumed. The front plane represents the axial-lateral plane of the transducer centered in elevation.

ments using a time-domain, explicit iterative solver. Validation of the model is accomplished using the analytic steady-state and transient solutions provided in (6) and (8). Additional validation of the transient solution of the explicit finite element code is achieved by comparison with implicit three-dimensional dynamic thermal finite element code that was written and implemented in the Finite Element Analysis Program (FEAP, Dr. Robert L. Taylor, Department of Civil and Environmental Engineering, University of California at Berkeley). Run times for the explicit solver are on the order of several hours using 1.67 GHz dual Athalon processing nodes, with RAM requirements of approximately 2.0 GB.

C. Single-Location ARFI Imaging

Tissue heating and cooling profiles are characterized for an ARFI interrogation at a single spatial location. A typical ARFI pulse sequence consists of six 28 µs high-intensity pushing pulses with 150 µs between each pulse, for an effective duty cycle of 15.7% and a total insonification time of 0.92 ms. This represents the timing of the high-intensity pushing pulses, unless otherwise specified. The models are presented for materials possessing different ultrasonic absorption coefficients ranging from 0.3–1.5 dB/cm/MHz.

D. Two-Dimensional ARFI Imaging

Tissue heating and cooling profiles are also characterized for ARFI interrogation of a two-dimensional ROI. As with color Doppler imaging, two-dimensional ARFI images are created by sequentially interrogating lateral locations. A typical ARFI ROI is comprised of up to 50 adjacent beam lines that are separated laterally by a minimum of 0.35 mm spatially spanning a total of 1.75 cm. Temporally, each interrogation is separated by 5 ms (i.e., time between initiation of the first pushing pulse in each lateral location) for a total ROI acquisition time of 0.25 seconds when 50 lines are interrogated. Simulation of two-dimensional ARFI heating is accomplished using convolution [20], [31]. The tissue heating associated with a single ARFI interrogation is characterized for 60 second sampling rate of 200 Hz. The tissue heating for two-dimensional ARFI imaging of the entire ROI is determined by spatially convolving the individual thermal results for each time step. Results are compared and contrasted for different beam spacings and two focal configurations (F/1.3 and an unfocused, narrow transmit aperture). Heating trends associated with real-time ARFI imaging are also simulated by repeatedly interrogating lateral locations at varying frame rates.

E. ARFI Shear Wave and M-mode Imaging

Radiation force imaging is also used for generating shear waves and tracking their propagation and interaction with structures in tissue [2], [7], [11], [23]. ARFI shear wave sequences involve repeated applications of high-intensity pulses in a given spatial location, while translating the tracking beam index across the desired field of view. Typically, the number of high-intensity interrogations that are applied at a given location is 20, each of which is separated by 5 ms during which displacement tracking occurs [11], [28]. The heating associated with ARFI shear wave sequences is characterized as a function of the total number of high-intensity interrogations at a given location. Heating associated with ARFI M-mode imaging can be viewed
as an extension of shear wave imaging, where ARFI interrogations also occur at a single location over time; however, the temporal spacing between successive interrogations is much greater (e.g., arterial wall imaging throughout the cardiac cycle).

F. Thermal Expansion

Coupled thermal-mechanical simulations are performed to predict the amount of tissue expansion that would occur due to tissue heating during ARFI imaging. The thermal expansion model is validated using the analytic expression in (9) to approximate the thermal expansion associated with the heating of a sphere.

G. Temperature-Dependent Changes in Sound Speed

Changes in sound speed due to tissue heating are simulated for breast and liver using ∂c/∂T values of −2.93 and 1.0 m/s/°C, respectively [25], [26]. Errors in displacement due to these changes in sound speed are calculated by determining the mean sound speed change from the transducer face to the focal depth along the center line (laterally) and solving for the error in estimated round-trip pulse-echo time. The error in round-trip time is then converted to a displacement error using the assumed constant sound speed of 1540 m/s.

IV. Results

A. Thermal Characterization of Single-Location ARFI Imaging

Fig. 2 shows the temperature increases in the axial-lateral plane (centered in elevation) for different tissue absorptions after a single ARFI interrogation. Note that the temperature scale, in degrees Celsius, is different for each tissue absorption.

Fig. 3(a) shows the increase in temperature at the focal depth of 2.0 cm during an ARFI pushing sequence for different absorptions. The bottom figure represents the corresponding timing diagram for the pushing pulses. The maximum temperature value that each simulation reaches at the focus is also the maximum temperature throughout the three-dimensional volume of tissue for that given absorption (Fig. 2). Fig. 3(b) illustrates the rate of cooling that occurs due to thermal conductivity at the focal point for absorptions ranging from 0.3–1.5 dB/cm/MHz. The cooling rate is the same for all tissue absorptions. The temperature at the focus has cooled to almost 40% of its peak value within 0.5 s, independent of the near field temperature distribution.

Fig. 4 shows cooling profiles at different depths as a function of different absorptions. Note that the temperature scales are different for each depth. These figures illustrate how cooling times are a function of both position relative to the focus and spatial heat distribution due to tissue absorption.

Fig. 5 shows the temperature increases along the axial-lateral plane (centered in elevation) for an unfocused, narrow transmit aperture configuration using 10 transmit elements for absorptions of 0.5 and 1.0 dB/cm/MHz. These temperature profiles are shown after a single ARFI interrogation (0.92 ms, 28 µs pushing pulses, duty cycle of 15.7%). These images can be contrasted with those in Fig. 2 (F/1.3 focal configuration).

B. Thermal Characterization of Two-dimensional ARFI Imaging

Fig. 6(a) and (b) shows the temperature increases after interrogation of a 1.75-cm ROI with an F/1.3 transducer configuration for tissue absorptions of 0.5 and 1.0 dB/cm/MHz, respectively. Fifty sequential ARFI interrogations are separated by 0.35 mm spatially and 5 ms temporally from right to left. Again, the temperature scale, in degrees Celsius, is different for these two cases. Note that the maximum temperature increase has shifted from the focal depth of 2.0 cm (the location of maximum temperature increase for a single ARFI interrogation, shown in Fig. 2), to 1.45 cm for an absorption of 0.5 dB/cm/MHz and to 0.2 cm for an absorption of 1.0 dB/cm/MHz. Both of these depths correspond to the location where the maximum spatial and thermal overlap occurs between sequential ARFI interrogation locations. Fig. 6(c) shows the maximum temperature generated in tissue as a function of the number of lateral positions separated by 0.35 mm spatially and 5 ms temporally. Note that as the ROI increases in size, the maximum temperature increase plateaus.

The heating associated with two-dimensional ARFI imaging is also modeled for an unfocused, narrow transmit aperture transducer configuration using 10 elements. The top row in Fig. 7 shows the temperature profile associated with interrogating a 1.75-cm ROI with this transducer configuration, where each adjacent interrogation line is separated by 0.35 mm spatially and 5 ms temporally. The bottom row of Fig. 7 shows a line spacing of 1.75 mm spatially, effectively eliminating spatial overlap between adjacent interrogations. Note the difference in the temperature scales between the top row and bottom row of figures.

For real-time ARFI imaging, the two-dimensional interrogations (shown in Figs. 6 and 7) will be repeated in time. Fig. 8 shows the temperature distribution for real-time ARFI imaging using an F/1.3 focal configuration for absorptions of 0.5 and 1.0 dB/cm/MHz. The 1.75-cm ROI has been imaged with a lateral spacing of 0.35 mm and a frame rate of 3 frames per second for a total of 3 seconds (i.e., the two-dimensional ROI is interrogated 8 times). Fig. 9 shows the corresponding real-time implementation for the unfocused, narrow transmit aperture configuration with a line spacing of 0.70 mm in the top row, and 1.75 mm in the bottom row.
Fig. 2. Temperature increases in the axial-lateral plane (centered in elevation) for different tissue absorptions after a single ARFI interrogation. The transducer is centered along the top of the image, spanning a total of 1.52 cm laterally. The focus is at 2.0 cm, with an F/1.3 focal configuration. Each figure uses a different temperature scale in degrees Celsius.

(a) $\alpha = 0.3$ dB/cm/MHz

(b) $\alpha = 0.5$ dB/cm/MHz

(c) $\alpha = 1.0$ dB/cm/MHz

(d) $\alpha = 1.5$ dB/cm/MHz

Fig. 3. Figure (a) shows heating that occurs at the focus during a single six-cycle ARFI interrogation with an F/1.3 focal configuration as a function of tissue absorption. The maximum temperature value that each simulation reaches at the focus is also the maximum temperature throughout the three-dimensional volume of tissue for that given absorption. The bottom of Figure (a) represents the corresponding timing diagram for the high-intensity pushing pulses. Figure (b) shows the normalized cooling that occurs at the focus for different absorptions immediately after an ARFI interrogation is complete. The cooling behavior at the focus is identical for all four absorptions. The temperature has cooled to almost 40% of its peak value at the focus within 0.5 s.
Fig. 4. Cooling over several seconds that occurs at different axial depths for different absorptions for an F/1.3 focal configuration. Near-field locations are laterally centered with respect to the transducer. Note the differences in temperature scales for each plot.

Fig. 5. Temperature increases along the axial-lateral plane (centered in elevation) for different absorptions. The transducer is centered along the top of the images, spanning a total of 0.2 cm. This is an unfocused, narrow transmit aperture configuration using 10 elements.

Fig. 6. Temperature increases after a single frame of two-dimensional ARFI imaging using an F/1.3 focal configuration, as shown in Fig. 2. In the simulation, 50 interrogated locations are laterally spaced 0.35 mm apart with a time delay of 5 ms between successive locations temporally. The figures on the left and in the middle demonstrate the thermal profiles for absorptions of 0.5 and 1.0 dB/cm/MHz, respectively. The temperature scales, in degrees Celsius, are different for each figure. Note that the maximum temperature increase has shifted from the focal depth of 2.0 cm (the location of maximum temperature increase for a single ARFI interrogation), to 1.45 cm for an absorption of 0.5 dB/cm/MHz and to 0.2 cm for an absorption of 1.0 dB/cm/MHz. The figure on the right shows how the maximum temperature varies as a function of the number of lateral positions in the ROI for this focal configuration, using the same spatial and temporal spacing.
Fig. 7. Temperature increases after a single frame of two-dimensional ARFI imaging using an unfocused, narrow transmit aperture configuration, as shown in Fig. 5. In the top row, 50 lines are laterally spaced 0.35 mm apart with a 5 ms time delay between successive locations temporally, whereas in the bottom row, 10 lines are laterally spaced 1.75 mm apart with a 5 ms time delay between successive lateral locations. The figures on the left demonstrate the thermal profile for an absorption of 0.5 dB/cm/MHz, and the figures on the right demonstrate the thermal profile for an absorption of 1.0 dB/cm/MHz. The temperature scales, in degrees Celsius, are different for each figure.

Fig. 8. Real-time, two-dimensional ARFI imaging, using an F/1.3 focal configuration, performed at a frame rate of 3 frames per second for a total duration of 3 seconds for absorptions of 0.5 and 1.0 dB/cm/MHz. The line spacing is 0.35 mm with a 5-ms temporal delay between successive locations. Note that the temperature scale, in degrees Celsius, is different for each figure. For an absorption of 0.5 dB/cm/MHz, this maximum temperature occurs at a depth of 1.45 cm, whereas for an absorption of 1.0 dB/cm/MHz, it occurs at a depth of 0.2 cm.
Fig. 9. Real-time two-dimensional ARFI imaging, using an unfocused, narrow transmit aperture configuration, performed at a frame rate of 3 frames per second for a total duration of 3 seconds for absorptions of 0.5 and 1.0 dB/cm/MHz. In the top row, the line spacing is 0.70 mm spatially and 5 ms temporally between successive locations. In the bottom row, the line spacing is 1.75 mm spatially and 5 ms temporally between successive locations. Note that the temperature scale, in degrees Celsius, is different for each figure.

Fig. 10. Maximum temperature increases generated by real-time ARFI imaging for an absorption of 0.5 dB/cm/MHz using an F/1.3 focal configuration for two frame rates and a variable number of lines. The line spacing is 0.35 mm between adjacent ARFI interrogations. There are 5 ms between sequential ARFI interrogations, with the time between frames being a function of the specified frame rate. These maximum temperatures occur at a depth of 1.45 cm. The 50-line simulation does not reach a steady-state temperature increase due to an artifact of the insulating boundary conditions that do not allow for energy to conduct beyond the finite volume of tissue being modeled.
The maximum temperature over the three-dimensional volume of tissue associated with real-time ARFI imaging using an F/1.3 focal configuration for an absorption of 0.5 dB/cm/MHz is characterized as a function of the number of lines per frame and the frame rate. Fig. 10 shows the maximum temperature (that occurs at a depth of 1.45 cm), for 1–50 lines, spaced 0.35 mm apart, for frame rates of 1 frame per second and 3 frames per second. For higher numbers of lines, steady-state temperature increases are not achieved due to an artifact of the insulating boundary conditions that do not allow for energy to conduct beyond the finite volume of tissue being modeled.

C. Thermal Characterization of ARFI Shear Wave and M-mode Imaging

Acoustic radiation force is being explored in the generation and imaging of shear waves in tissue [2], [7], [11], which in our implementation involves repeating interrogations at the same spatial location, while sequentially tracking shear wave propagation at different lateral locations [2], [11]. Figs. 11(a) illustrates a linear increase in the maximum temperature, which occurs at the focus, as a function of the number of ARFI interrogations that are applied at a given spatial location over time during ARFI shear wave imaging. Fig. 11(b) shows the maximum temperature increase, which occurs at the focus, in ARFI M-mode imaging for each absorption through time. ARFI M-mode imaging is analogous to shear wave imaging except that the time delay between successive interrogations is extended to 50 ms, instead of 5 ms for shear wave imaging.

Fig. 11(c) and (d) illustrates the temperature increase associated with ARFI M-mode imaging for an F/1.3 focal configuration for absorptions of 0.5 and 1.0 dB/cm/MHz, respectively. These temperature increases occur over 7 seconds while an ARFI interrogation is being fired every 50 ms (for a total of 140 ARFI interrogations). Note that the temperature scales, in degrees Celsius, are different for the two figures.

D. Thermal Expansion

A coupled thermal-structural simulation for an absorption of 0.5 dB/cm/MHz is implemented to determine how much thermal expansion occurs during ARFI imaging. The coupled simulation is implemented for the same thermal configuration illustrated in Fig. 2(b). Fig. 12(a) shows the magnitude of the maximum displacement (in microns) from thermal expansion that occurs immediately after completion of the ARFI interrogation. The peak magnitude of displacement is 0.055 μm, but the greatest axial component of displacement is less than 0.02 μm.

E. Temperature-Dependent Changes in Sound Speed

Fig. 12(b) and (c) shows the change in sound speed and associated displacement errors in breast fat and liver as a function of axial position along the center line of the transducer (where maximum heating occurs) for a single ARFI interrogation for an absorption of 0.5 dB/cm/MHz. Note that the displacement errors accumulate with axial depth away from the transducer.

V. Discussion

A. Thermal Characterization of a Single ARFI Interrogation

Temperature distributions arising from ultrasonic energy absorption depend on tissue properties. The absence of blood perfusion effects in these models creates maximum heating conditions and worst-case estimates of tissue temperature increases. The absorption of soft tissue varies significantly. Since absorption is a frequency-dependent parameter, modeling a range of absorptions also provides insight into performing ARFI imaging through a range of frequencies in soft tissue.

For low-to-moderate tissue absorptions (0.3–0.5 dB/cm/MHz), there is a significant focal gain in intensity, and the majority of energy absorption and heating occurs near the focus (Fig. 2). More absorbing tissues (1.0–1.5 dB/cm/MHz), such as the breast [32], absorb energy over a much larger volume in the near field and generate smaller temperature increases. This is demonstrated by the order of magnitude difference in the maximum temperature increase for tissue absorptions that have been modeled (0.3–1.5 dB/cm/MHz, shown in Fig. 2). For all tissue absorptions modeled with an F/1.3 configuration, however, the maximum temperature for a single ARFI interrogation still occurs at the focus. This is consistent with the findings of Herman and Harris [18]. For a single ARFI interrogation, transducer heating is assumed to be negligible because the total duration of insonification is less than 1 ms.

As Fig. 3 illustrates, the time between the six high-intensity pushing pulses in a single ARFI interrogation is too short for appreciable cooling to occur in the tissue. Therefore, the heating during a single ARFI interrogation is directly related to the total number of high-intensity cycles within a few milliseconds. For example, a 100% duty cycle 168-μs pulse would result in the same temperature increase as six 28-μs pulses transmitted at a 15.7% duty cycle in 0.92 ms. For ARFI imaging, this finding is important because tissue demonstrates larger maximum displacements for a 100% duty cycle as compared to a 15.7% duty cycle, and there is no heating penalty for using a longer, single pulse.

Cooling effects become appreciable on the order of tenths of a second at the focus, with the maximum focal temperature decaying to 50% of its peak value in less than 0.5 seconds [Fig. 3(b)]. Fig. 3(b) also demonstrates that focal point cooling times are independent of tissue absorption for an F/1.3 focal configuration. This is due to the similarity in relative heat distribution that occurs at the focus for all tissue absorptions with this focal config-
Fig. 11. Figure (a) shows a linear increase in the maximum temperature as a function of the number of ARFI interrogations during ARFI shear wave imaging when interrogations are fired every 5 ms (covering a span of 175 ms). Figure (b) shows the maximum temperature increase, which occurs at the focus, during ARFI M-mode imaging for absorptions of 0.5 and 1.0 dB/cm/MHz, with 50 ms between ARFI interrogations. Figures (c) and (d) illustrate the temperature increases associated with ARFI M-mode imaging for 7 seconds while interrogating every 50 ms (for a total of 140 interrogations) for absorptions of 0.5 and 1.0 dB/cm/MHz respectively. All of these models are for an F/1.3 focal configuration.

Fig. 12. Figure (a) shows the magnitude of the displacement due to thermal expansion immediately after an ARFI interrogation for an absorption of 0.5 dB/cm/MHz. Displacements are in microns. The peak magnitude of displacement is 0.055 \( \mu \)m, but the greatest axial component of displacement is less than 0.02 \( \mu \)m. Figure (b) illustrates the change in sound speed in breast fat and liver as a function of axial position along the center line of the transducer (where maximum heating occurs) for an absorption of 0.5 dB/cm/MHz after a single ARFI interrogation. Changes in sound speed are in meters per second. Figure (c) shows how these changes in sound speed would be reflected as displacement errors. Notice that these errors accumulate with axial distance from the transducer. Negative changes in displacement would result in underestimation of radiation force-induced displacements.
uration (Fig. 2), though maximum temperatures vary due to the total amount of energy delivered to the focal region. The cooling behavior of tissue is more complicated at locations other than the focal point because the distribution of thermal energy is a function of tissue absorption. As demonstrated in Fig. 4, there is a complex relationship between the magnitude and the spatial distribution of heat in the near field. At a depth of 1 cm above the focus, temperatures will increase for approximately 1 second as heat from the focal region dissipates through the surrounding tissue. Whereas the magnitude of these temperatures is significantly lower than those experienced at the focus, this heat will accumulate when two-dimensional and real-time ARFI imaging is performed, due to the overlapping thermal profiles in the near field between successive interrogations. This explains why the peak temperature for two-dimensional ARFI occurs at a different depth than for a single interrogation.

Fig. 5 illustrates the temperatures increases from an unfocused, narrow transmit aperture configuration (10 elements) to compare with the F/1.3 temperature distributions shown in Fig. 2. As is expected, a more uniform distribution of heat deposition is present as compared with the F/1.3 focal configuration. Again, for higher tissue absorptions, more heat is deposited in the near-field. While the maximum temperatures are similar to those for the F/1.3 focal configuration (with a peak in situ intensity value of $I = 1,000 \text{ W/cm}^2$ used for both configurations), the lateral extent of the heating field is reduced due to the smaller size of the active transmit aperture.

### B. Thermal Characterization of Two-dimensional ARFI Imaging

To achieve two-dimensional ARFI imaging, multiple interrogations are fired in close temporal (5 ms) and spatial (0.35–1.75 mm) proximity to one another. As Fig. 6 illustrates, the thermal effects of interrogating multiple spatial locations are not directly cumulative. For higher tissue absorptions, the maximum temperature increase that occurs is located in the near field (e.g., at a depth of 0.2 cm from the tissue surface for $\alpha = 1.0 \text{ dB/cm/MHz}$). The near-field location of this peak heating is a function of the lateral distribution of heat in an interrogation and the spacing of adjacent interrogations. Therefore, it is important to monitor heating at locations other than the focal depth for two-dimensional ARFI imaging, and these models can be used to determine the locations of maximum heating through time for different imaging implementations.

By using a smaller aperture (Fig. 5) in more absorbing tissues, there is a reduction in the lateral spatial overlap of thermal profiles, and the resulting cumulative heating is reduced (Fig. 7). For lower tissue absorptions, the temperature effects remain confined to the focal region because there is less near-field heating and less overlap of adjacent heating profiles. Of course, use of a lower ultrasound frequency in the more absorbing tissues would also reduce near-field heating.

As Fig. 6(c) illustrates, for large enough ROIs, a plateau in maximum temperature is reached as the cooling effects of early interrogations balance the heating effects of later interrogations. For the higher tissue absorptions, it takes a larger ROI for this plateau to be reached because the lateral extent of the near-field heating is greater than with lower tissue absorptions. These plateaus in heating also represent when the location of the maximum heating has reached its steady-state location in the near-field.

Reduction of the temperature increase during two-dimensional ARFI imaging can be achieved by increasing the spacing between adjacent interrogation locations. This is most clearly demonstrated with the unfocused transducer configuration in which the lateral extent of heating at a single location is uniform through all tissue depths (Fig. 5). Fig. 7 illustrates how an increase in the spacing can minimize the heating between adjacent interrogation locations and reduce the maximum temperature increase associated with two-dimensional ARFI imaging.

For real-time implementation of two-dimensional ARFI imaging, a trade-off exists between frame rate (Fig. 10), spacing of adjacent lines (Fig. 9), and the extent of the ROI (Fig. 6). The general trends shown in Figs. 9 and 10 demonstrate that decreasing the frame rate and increasing the spacing between adjacent lines lowers the steady-state temperature that would be achieved in real-time ARFI imaging. Maintaining temperature increases of less than 6°C is possible; however, the frame rate, lateral resolution, and duration over which real-time ARFI imaging is performed must be best selected accordingly.

### C. ARFI Shear Wave and M-mode Imaging

For ARFI shear wave imaging, the benefit of spatial offset for cooling between successive ARFI interrogations is lost since multiple shear waves must be generated at a single spatial location (in the absence of parallel processing for tracking). Cooling, therefore, occurs only during the temporal delay between successive ARFI interrogations. Tissues experience a greater thermal burden during ARFI shear wave imaging, especially for lower tissue absorptions where thermal energy is concentrated in smaller volumes of tissue. As would be expected, the distribution of heat for shear wave imaging is similar to that for single ARFI interrogations for a given tissue absorption (Fig. 2). The thermal burden that the tissue experiences in shear wave imaging would be greatly reduced if an ARFI imaging scanner were capable of parallel receive processing, where shear waves generated from a single ARFI interrogation could be tracked across an entire field of view [7]. Multiple instances of shear wave generation at the same spatial location would not be necessary, and temperature increases would be equal to those for a single ARFI interrogation (Fig. 2).

ARFI M-mode imaging is used in gated interrogation of blood vessels throughout the cardiac cycle. ARFI M-mode imaging is analogous to shear wave imaging; however, much longer delays occur between interrogations.
For ARFI M-mode imaging, the temperature increase is related to the amount of time between successive ARFI interrogations and the amount of heat lost to blood perfusion (which is not included in these models), especially when imaging blood vessels.

D. Thermal Effects on Displacement Tracking

ARFI imaging uses correlation-based methods to track tissue displacements as small as 0.2 μm over 0.3-mm search windows with approximate volumes of 0.072 mm³ (0.3 mm × 0.8 mm × 0.3 mm). The displacements due to thermal expansion are on the order of hundredths of a micron for a single ARFI interrogation [Fig. 12(a)]. The majority of this displacement occurs in the lateral direction, and will not impact ARFI displacement measurements because a new reference line is transmitted every 5 ms for all modes of ARFI imaging. Therefore, cumulative displacements due to thermal expansion over long periods of time are not a concern.

Over- and underestimation of displacement by a few microns can occur due to changes in sound speed from ARFI-generated tissue heating [Fig. 12(c)]. However, experimental data to date do not appear to demonstrate appreciable heating artifacts. This could be related to an overestimation of ∂T/∂c or, more likely, an overestimation of ARFI in situ intensities determined using linear extrapolation of small-signal derated fields [30].

Changes in sound speed in the lens will also affect displacement estimates. These artifacts would appear as constant displacement offsets at all depths as opposed to the depth dependent accumulation of displacement shown in Fig. 12(c). Heating artifacts can be corrected using filters in ARFI image processing algorithms.

E. Limitations of the Thermal Model

1. Mechanical Waves: The models that have been presented have assumed that all energy that is absorbed at a given location will lead to tissue heating at that location. However, with the transfer of momentum associated with radiation force, strain energy is stored in the radiation-force induced tissue displacement fields. After the high-intensity acoustic pulses are applied, the tissue relaxes, and shear and dilatational displacement waves propagate away from the focal region. As these waves disperse, they are attenuated by thermo-viscous mechanisms. Therefore, this mechanical energy is converted to thermal energy, but for materials with low thermo-viscous attenuation, this is not in the focal region. Therefore, the thermal profiles generated by ARFI imaging should be considered functions of tissue stiffness and viscosity, properties that affect mechanical wave velocity and absorption, but these parameters have not been incorporated into these models. These effects will be addressed in a future paper.

2. Nonlinear Propagation: Nonlinear propagation, and its effects on acoustic radiation force, has been extensively treated in the literature [2], [30], [33]–[35]. For the same temporal average intensity, a wave with higher pressure amplitude and shorter pulse duration generates a larger radiation force than does a lower amplitude, longer duration acoustic wave. This larger radiation force is due to the higher-order harmonics generated by nonlinear propagation, which result in an increase in energy absorption related to a finite amplitude absorption parameter [2], [30], [33], [34], [36]. In addition to increasing the force magnitude, nonlinear propagation is generally associated with a translation of the location of the peak in the force field, and therefore the heating, closer to the transducer. However, this effect decreases with increasing center frequency and finite amplitude absorption that exists in soft tissue as compared to aqueous solutions [30]. ARFI imaging at a center frequency of 7.2 MHz in soft tissues provides an effectively high tissue absorption, which decreases the described effects of nonlinear propagation, and instead results in a reduction of the linearly estimated focal intensities [30]. Therefore, linear models of tissue intensity fields provide leading order estimates of ARFI imaging tissue heating, which can be refined using nonlinear propagation corrections [30]. These corrections will be more significant for applications of ARFI imaging at greater tissue depths with lower ultrasonic frequencies. For these implementations, nonlinear simulation packages with finite amplitude absorption parameters will need to be utilized to model the tissue intensity fields.

3. Transducer Heating: In addition to the temperature increases created by absorption of the ultrasound beam in radiation force imaging, transducer heating may become a concern, but has not been included in these models. While the transducer heating associated with a single ARFI interrogation is assumed to be negligible due to the short insonification time, the cumulative insonification time for two-dimensional real-time and shear wave ARFI imaging warrants further study. Transducer heat can appreciably conduct to two centimeters below the tissue surface during continuous imaging [22], and will have a cumulative effect when coupled with heating due to tissue absorption.

4. Finite Tissue Volume: These models simulate a finite volume of tissue with insulating boundaries that do not allow for thermal energy to escape into the environment or surrounding structures later in time. These boundary conditions generate artifacts in the long-duration, real-time simulations in which heat has time to appreciably conduct to the model boundaries. The models also do not incorporate a radiation boundary at the tissue surface, providing only a conductive mechanism for heat transfer in the lateral and elevation dimensions in the near field. This, again, may artificially increase the temperature in the near-field, especially in the real-time models for high tissue absorptions in which the maximum heating occurs within a centimeter of the surface. All of the implemented boundary conditions simulate worst-case heating that could occur during ARFI imaging.
**F. Applications to Other Radiation Force-Based Imaging Modalities**

The models and results that have been presented in this paper can be easily extended to the heating associated with other radiation force-based imaging modalities due to energy absorption by soft tissue. Modalities that use lower center frequencies will need to take into account the effects of nonlinear propagation, and modalities that involve contrast agents or imaging of tissue-bone interfaces will need to include heating effects associated with reflective sources.

**VI. CONCLUSION**

Acoustic radiation force-based imaging methods appear to have clinical potential for characterizing the local mechanical properties of tissue. The models presented in this paper characterize the heating that is associated with ARFI imaging as currently implemented as a function of different soft tissue absorptions and focal configurations. The temperature rise at the focus is greater for less absorbing tissues, whereas more absorbing tissues distribute the thermal energy over greater volumes in the near field. By adjusting focal configurations and/or frequency, near-field heating and overlap of adjacent beams can be controlled. Creation of two-dimensional ARFI images is not limited by thermal effects because sufficient cooling occurs spatially and temporally between adjacent ARFI interrogations. Frame rate, lateral resolution, size of the region of interest, and duration of real-time, two-dimensional ARFI imaging are parameters that can be optimized in future development of ARFI beam sequences to control tissue heating. The thermal burden associated with ARFI shear wave imaging can be significantly reduced with parallel receive processing. The effects of thermal expansion are negligible in ARFI image reconstruction, but ARFI images can suffer from artifacts due to sound speed changes in the transducer and the tissue.

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**REFERENCES**


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