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Acoustic radiation force impulse imaging of vulnerable plaques: a finite element method parametric analysis

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Abstract

Plaque rupture is the most common cause of complications such as stroke and coronary heart failure. Recent histopathological evidence suggests that several plaque features, including a large lipid core and a thin fibrous cap, are associated with plaques most at risk for rupture. Acoustic Radiation Force Impulse (ARFI) imaging, a recently developed ultrasound-based elasticity imaging technique, shows promise for imaging these features noninvasively. Clinically, this could be used to distinguish vulnerable plaques, for which surgical intervention may be required, from those less prone to rupture. In this study, a parametric analysis using Finite-Element Method (FEM) models was performed to simulate ARFI imaging of five different carotid artery plaques across a wide range of material properties. It was demonstrated that ARFI could resolve the softer lipid pool from the surrounding, stiffer media and fibrous cap and was most dependent upon the stiffness of the lipid pool component. Stress concentrations due to an ARFI excitation were located in the media and fibrous cap components. In all cases, the maximum Von Mises stress was < 1.2 kPa. In comparing these results with others investigating plaque rupture, it is concluded that while the mechanisms may be different, the Von Mises stresses imposed by ARFI are orders of magnitude lower than the stresses associated with blood pressure.

Keywords

Acoustic Radiation Force Impulse (ARFI); Medical imaging; Atherosclerosis; Plaque rupture; Elasticity

1. Introduction

Cardiovascular disease is the leading cause of death and disability in the United States (Roger et al., 2011). The majority of cardiovascular clinical events are attributed to atherosclerosis, a process in which fatty deposits, inflammation, cells, and scar tissue accumulate between the endothelium lining and smooth muscle walls of arteries (Naghavi,

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Conflicts of interest statement

Authors M.L. Palmeri and G.E. Trahey are intellectual property owners of Acoustic Radiation Force Impulse (ARFI) imaging.

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2010). This can result in the formation of plaques, whose rupture has been shown to be the most common cause of complications including sudden cardiac death, acute coronary syndromes, and/or stroke (Casscells et al., 2003; Davies, 2000; Falk et al., 1995; Virmani et al., 2000).

It is widely accepted that plaques with a large, soft, lipid-rich necrotic core covered by a thin fibrous cap are more vulnerable to rupture (Falk et al., 1995; Lee and Libby, 1997). Plaque rupture is believed to occur when the fatigue strength of the fibrous cap is exceeded. This has been supported by several groups who, using Finite Element Analysis (FEA) techniques, reported increased stresses in ruptured plaques compared to those with no evidence of rupture found in histology (Cheng et al., 1993; Huang et al., 2001; Li et al., 2006). The presence of a lipid pool is believed to be particularly dangerous because it confers greater stress within the overlying cap and hence increases the propensity of rupture (Loree et al., 1994). For this reason, and also because formation of the lipid pool precedes that of the fibrous cap and ultimately plaque rupture, the ability to reliably differentiate a soft lipid core from an otherwise stiffer, more stable region may help clinicians in treatment planning.

Recent developments in medical imaging have allowed for the morphological and structural characterization of plaques *in vivo* (Sanz and Fayad, 2008; Waxman et al., 2006). Using high resolution magnetic resonance imaging (MRI), the lipid, fibrous, calcified, hemorrhagic, and thrombotic regions in plaque tissue have been differentiated using histology as a gold standard (Hatsukami et al., 2000; Hatsukami and Yuan, 2010; Kawahara et al., 2007; Saam et al., 2007; Touze et al., 2007; Trivedi et al., 2004). Catheter-based probes using optical coherence tomography (OCT) (Tearney et al., 2006) or intravascular ultrasound (IVUS) provide information regarding plaque size and arterial wall thickness in the coronary artery. For differentiating soft lipid cores from more stable calcified regions, an alternative to the high cost associated with MRI and the invasive nature of catheter-based techniques is a recently developed ultrasound-based elasticity imaging method, termed Acoustic Radiation Force Impulse (ARFI) imaging.

ARFI imaging characterizes the mechanical properties (i.e. stiffness) of soft tissue by measuring the tissue deformation from a localized acoustic radiation force excitation (Nightingale et al., 2001). Briefly, the method uses a single ultrasound transducer to 1) apply an impulse-like (short duration) excitation within the tissue and 2) track the resulting deformation spatially and temporally using ultrasonic correlation-based methods. ARFI generates sufficient acoustic radiation force (on the order of dynes) capable of displacing soft tissue approximately 1–10 μm (Nightingale, 2011). The axial displacements are inversely related to tissue stiffness; meaning that soft regions will displace more than stiffer regions (Palmeri et al., 2006a; Palmeri et al., 2005). Similar to diagnostic B-mode imaging, the interrogation region can be translated across the transducer to create 2-D tomographic images that reflect the stiffness of soft tissue. For a thorough description of ARFI imaging and the various clinical settings in which the method has been applied, the reader is referred to Doherty et al. (Doherty et al., 2012). To characterize plaque and arterial tissues, the method has been used in several *ex vivo* and *in vivo* scenarios (Dumont et al., 2006; Dumont et al., 2009; Trahey et al., 2004). Most recently, Dahl et al. (Dahl et al., 2009) demonstrated the use of ARFI imaging *in vivo* to differentiate suspected soft plaques from hard plaques in the carotid artery.

To investigate the effects of material properties on the ability of ARFI to differentiate a soft lipid pool from an otherwise stiffer region, this study uses Finite-Element Method (FEM) models to simulate ARFI imaging of carotid artery plaques. A parametric analysis is performed in five different plaque geometries by varying the specified material properties.

Because fracture of a fibrous cap could lead to significant consequences, the Von Mises stress contours induced by an ARFI excitation are also investigated.

2. Methods

The FEM models used herein are adapted from models previously developed by Palmeri et al. (Palmeri et al., 2005) for investigating the response of soft tissue to an ARFI excitation. These models have been experimentally validated in calibrated gelatin-based tissue-mimicking phantoms (Nightingale et al., 2006; Palmeri et al., 2005) and used to investigate the dynamic response of elastic inclusions to an ARFI excitation (Palmeri et al., 2006a).

2.1 Plaque Mesh Generation

Three-dimensional, rectangular, solid meshes consisting of 0.1mm, trilinear, cubic elements were created using LS-PREPOST2 (Livermore Software Technology Corp., Livermore, CA). The mesh extended ± 5 mm in the lateral (x) dimension, 5mm in the elevation (y) dimension, and 25mm in the axial (z) dimension. Half-symmetry was assumed on the front face (x - z plane, $y = 0$ mm). The back (x - z plane, $y = 5$ mm), left (y - z plane, $x = -5$ mm), and right (y - z plane, $x = 5$ mm) faces were assumed to be non-reflecting. Full-constraints were assumed for the top (x - y plane, $z = 1$ mm) and bottom (x - y plane, $z = 26$ mm) faces. No slip was allowed at material boundaries. Five carotid artery plaque geometries were constructed based on published histology and *in vivo* MRI images by Li et al. (Li et al., 2006). Each image was manually segmented to define the media (i.e. arterial wall), lumen, fibrous cap, and lipid pool components and positioned in a tissue component. Material properties were assigned using custom-written MATLABTM (MathWorks, Natick, MA) scripts. The geometry of the lateral-axial plane was assumed constant in the elevation dimension to construct 3-D models from the 2-D segmented images.

2.2 Material Properties

Literature on the material properties of human arteries and is not extensive and varies considerably (Hoskins, 2007). Choosing values consistent with the literature (Baldewsing et al., 2004; Baldewsing et al., 2005; Cheng et al., 1993; de Korte et al., 2000; Lee et al., 1996; Loree et al., 1992) a model with $E_{media} = 1000$ kPa, $E_{cap} = 1000$ kPa, and $E_{lipid} = 25$ kPa was defined as a baseline case; where E represents the static Young's Modulus. A parametric analysis was conducted by varying the Young's modulus of each component according to the modified values listed in Table 1. The background tissue was modeled with $E_{tissue} = 4$ kPa in all simulations. The material was modeled as a linear, isotropic, and elastic solid with a Poisson's ratio of 0.495.

Arterial tissues are known to exhibit nonlinear behavior throughout the cardiac cycle (Kamenskiy et al., 2012). However, the quasi-linear assumptions herein are appropriate for the infinitely small strains ($<1\%$) that are typically encountered with ARFI imaging as validated by the models in this study. Also, while appreciable changes in blood pressure occur over the entire cardiac cycle, it is reasonable to expect that changes occurring over the short duration (< 50 msec.) used to acquire ARFI imaging data would be quite small, such that linear assumptions would be valid. Additionally, changes in arterial stiffness that may be attributed to nonlinear behavior are within the wide range of stiffnesses considered in the parametric analysis performed herein.

2.3 Acoustic Radiation Force Excitation

Using Field II (Jensen, 1996; Jensen and Svendsen, 1992), a linear acoustic field simulation package, the pressure fields were simulated for a linear cardiovascular transducer array according to the parameters listed in Table 2, where the transmit $F\#$ describes the focal

geometry of the transmitted beam. These imaging parameters have been demonstrated to be effective for ARFI imaging of carotid plaques, while maintaining FDA mechanical index and thermal index safety guidelines (Doherty et al., 2011).

Simulated pressure fields were converted to intensities and scaled according to empirically determined intensity values obtained from hydrophone measurements recorded using a transducer with the same parameters listed in Table 2. The acoustic radiation force field ($\vec{F}$) was determined according to Equation 1 (Doherty et al., 2012; Palmeri et al., 2005) where (I) is the pulse average intensity field, α is the tissue absorption coefficient, and c is the speed of sound.

$$\vec{F} = \frac{2\alpha}{c} \vec{I} \quad \text{Equation 1}$$

In soft tissues, the absorption coefficient is approximated by the tissue attenuation coefficient (Christensen, 1988). Attenuation, which is unknown and varies *in vivo*, modulates both the magnitude and also the geometry of the applied force (Doherty et al., 2012). To investigate these effects, the simulated acoustic radiation force was modeled with varying degrees of attenuation for values of $\alpha = 0.3$ and $\alpha = 0.7$, with a baseline value of $\alpha = 0.5$ dB/cm/MHz.

2.4 Computational Modeling & Data Analysis

The FEM simulations were performed using an explicit time-domain finite element algorithm available in LS-DYNA (Livermore Software Technology Corp., Livermore, CA). Effects of varied Poisson ratio and mesh dependencies (i.e. element size) were investigated. The values used herein were chosen because they minimized simulation runtime and provided results nearly identical to the convergent solution. For each simulation, datasets describing the axial displacements and associated Von Mises stresses were obtained. The Von Mises stress was chosen to represent an equivalent stress induced by the ARFI excitation. The results depicted herein represent the response that is co-planar with that of the maximum applied force.

Figure 1 depicts the simulated axial displacements in a homogeneous, isotropic, and elastic solid material with a Young's modulus of 4 kPa induced by an ARFI excitation. The dynamic response portrayed in Figure 1a describes the material deformation occurring at a single lateral location as a function of axial depth and time following excitation. The white dashed line depicts the time at which the maximum axial displacement occurs. Figure 1b shows the maximum axial displacements through time (i.e. occurring at the time indicated by the white dashed line in Figure 1b) as a function of lateral and axial position from a single excitation centered laterally. To simulate 2-D ARFI imaging in which the applied force (push) location is segmentally translated across the lateral dimension, the dynamic tissue response was modeled in separate simulations for each of nine lateral push locations spaced 1.2mm apart. Figure 1c is the final synthesized dataset that was created by combining these responses.

The magnitude of the applied acoustic radiation force can vary significantly, owing to many factors such as geometry, attenuation, scan angle, and imaging depth (Doherty et al., 2012). Therefore, with ARFI imaging it is the relative, and not the absolute, displacements that are most important. For comparing the results obtained herein, the ability of ARFI imaging to differentiate a soft lipid pool from a stiffer, more calcified region with a composition similar to that of the media and fibrous cap is quantified by computing the contrast between the simulated displacements in these regions. As defined according to Equation 2, the contrast is

the ratio of the mean axial displacement in the lipid pool $\bar{\delta}$ to that of the combined media and fibrous cap regions $\bar{\delta}_{media/cap}$, where the axial displacements (δ) correspond to the maximum displacement occurring through time.

$$Contrast = \frac{\bar{\delta}_{lipid}}{\bar{\delta}_{media/cap}} \quad \text{Equation 2}$$

3. Results

The maximum axial displacements and Von Mises stresses induced by an ARFI excitation were determined for five carotid artery plaque geometries as a function of specified material properties. Figure 2 shows the results obtained for the baseline material parameters. The images in Figure 2a represent the generated FEM meshes, depicting the segmentation of the various plaque components. The maximum axial displacement images shown in Figure 2b represent the synthesized 2-D ARFI images. Softer lipid pools displace more than the surrounding stiffer fibrous cap and media regions. Note that vertical bands of increased displacements observed in Figure 2b were also observed in the homogeneous example portrayed in Figure 1c. These bands are due to the spacing between adjacent acoustic radiation force excitations distributed across the lateral dimension. The maximum Von Mises stress contours shown in Figure 2c indicate stress concentrations localized within the thin regions of the fibrous cap and media. Vertical bands of increased stress, most clearly observed in the thick media portion of the proximal arterial wall in Model E, occur at the edges of the applied excitations.

In comparing the results among the investigated models, there is great variability in the maximum axial displacements shown in Figure 2b. For instance, the lipid component in Model D has a mean axial displacement of 2.62 ± 0.70 compared with 1.22 ± 0.33 μm in Model B. The lipid displacement also varies within a given model, a result most clearly observed in Model C. There is also great variability in the Von Mises stress portrayed in Figure 2c for the different models.

The baseline results represent the modeled response for a specific set of material properties. To better understand the contribution of each component, the following results compare the individual effects of tissue attenuation (Figure 3), media stiffness (Figure 4), fibrous cap stiffness (Figure 5), and lipid stiffness (Figure 6). In these figures, a single model geometry that demonstrates the common trends observed among all models is portrayed for conciseness. The contrast (Equation 2) between the lipid pool and the combined media and fibrous cap regions, along with the maximum Von Mises stress, are portrayed in Figure 7 for all model geometries and simulated parameters.

Increased attenuation resulted in a slight increase in the axial displacements of all components, as demonstrated by Model D in Figure 3. While Figure 7a shows a slight decrease in contrast between the lipid pool and the surrounding media and fibrous cap components with increased attenuation, qualitatively the effects portrayed in Figure 3 appear rather insignificant. Larger Von Mises stresses were observed with increased attenuation, concentrating in the thin regions of the media and fibrous cap components.

In Figure 4, a decreased media stiffness in Model B is associated with an increase in the axial displacements of the media component, but has little effect on the displacements of the lipid and fibrous cap components. This results in a loss of contrast portrayed in Figure 7b. Qualitatively it is observed in Figure 4 that as the stiffness of the media approaches that of the softer lipid component, it is more difficult to distinguish the boundaries of the lipid

region. However, even for values of $E_{media} = 200$ kPa the mean lipid displacement is still greater than that of the stiffer media and fibrous cap for all cases. In all modeled geometries, the Von Mises stress within the media component increased with increasing media stiffness. In all but Model D, this was associated with a decreased stress within the fibrous cap. Differences in the maximum Von Mises stress location account for the different trends observed in Figure 7b. In Figure 4, note that as the media stiffness increases from $E_{media} = 200$ kPa in Model B, the location of maximum Von Mises stress shifts from the fibrous cap to the media, typically at the proximal wall. For cases where the media component is extremely thin, the increased Von Mises stress can be quite significant. For instance, as observed in Figure 7b, the Von Mises stress in Model B changes from 667 Pa with $E_{media} = 200$ kPa to 1142 Pa with $E_{media} = 2000$ kPa.

Demonstrated by Model E in Figure 5, the axial displacements of the fibrous cap decreased with increasing stiffness of the fibrous cap. While the percent change in mean axial displacement within the fibrous cap between $E_{cap} = 1000$ kPa and $E_{cap} = 5000$ kPa can be significant, for instance 51 % in Model B, the overall effect of fibrous cap stiffness on lipid pool detectability in Figure 5 appears small. In Figure 5, note that increased cap stiffness is associated with an increase in the magnitude and also the number of stress concentrations. The maximum Von Mises stress in each model across all modified parameters was obtained for $E_{cap} = 5000$ kPa, with a maximum value of 1154 Pa observed in Model D (Figure 7c).

For each model geometry, with decreased lipid stiffness the simulated axial displacements within the lipid region increased, resulting in improved visualization of the lipid component, as demonstrated in Figure 6 for Model A. The contrast results in Figure 7d indicate contrast values >4.2 were obtained for all models with $E_{lipid} = 0.5$ kPa. With a value of $E_{lipid} = 50$ kPa, the contrast values are <1.7 for all models and as shown in Figure 6, it can be very difficult to visualize the lipid pool compared to the surrounding media and fibrous cap for large lipid stiffness values. There was no appreciable difference in either the location or magnitude of the maximum Von Mises stresses for the considered lipid stiffness values, as shown in Figure 6 and Figure 7d for all models.

4. Discussion

In this work, FEM analysis techniques were used to simulate ARFI imaging of carotid artery plaques. The ability of ARFI imaging to visualize plaques with soft lipid filled cores, believed to be more vulnerable than stiffer, calcified regions, was demonstrated across a wide range of material properties for five different geometries. In each case, the stresses associated with the ARFI excitation, where fracture of a fibrous cap could lead to significant consequences, were also evaluated.

4.1 Plaque Characterization

An appreciable increase in contrast was observed for low lipid stiffness and/or high fibrous cap and media stiffness, suggesting that ARFI imaging is best suited for imaging the most vulnerable of plaques, where increased ratios of the lipid to fibrous cap stiffness have correlated with evidence of rupture (Li et al., 2006). Changes in contrast were most dependent upon the material properties of the lipid region. While consistency in the trends observed for all geometries reported herein suggests true dependencies in the material properties, differences observed between the models are likely functions of geometry as well. The results for the baseline material parameters in Figure 2b support this hypothesis, however, a more detailed investigation using simplified models with controlled geometries is warranted.

The simulated results in this study corroborate those from ARFI imaging of excised arteries, in which areas of increased displacements corresponded to soft lipid filled regions found in histology (Trahey et al., 2004). However, factors including sources of error inherent to ultrasonic measurement techniques can complicate the results reported in these simulations (McAleavey et al., 2003; Palmeri et al., 2006b; Palmeri et al., 2005). In simulated ARFI tracking experiments, Pinton et al. (Pinton et al., 2006) reported a theoretical lower limit of the jitter, the standard deviation of the estimate error, equal to $1.05 \mu\text{m}$ for a typical SNR of 30 dB. With mean axial displacements often $< 2 \mu\text{m}$ observed in the media and fibrous cap, it is expected that the measured contrast presented in these simulations overestimate what can be achieved experimentally for the same material properties.

4.2 Stress & Safety Considerations

To our knowledge, this work presents the first study investigating the resulting stresses in arterial plaques due to an ARFI excitation. The magnitude, and in some cases the location, of the maximum Von Mises stress was observed to depend upon the stiffness of the plaque components. In all cases, the maximum Von Mises stress significantly increased with increasing fibrous cap stiffness (Figure 7c). It was observed that stress concentrations could shift from locations in the fibrous cap to those in the media with increased media stiffness (Figure 4). It is unlikely, however, that the stiffness of the fibrous cap would be less than that of the media. A negligible effect was observed for changes in lipid stiffness on the maximum Von Mises stress (Figure 7d). This is inconsistent with the results of others, who found that increased lipid pool stiffness resulted in a reduced peak stress due to blood pressure (Li et al., 2006; Loree et al., 1994). The difference here is likely due to the fact that with blood pressure, the fibrous cap is adjacent to the source of the imposed load such that the lipid can bear some of the stress. However, as observed with ARFI imaging, stresses also concentrate throughout the media near the tissue interface, where the lipid has less supporting role.

To provide perspective on the magnitude of the stresses associated with ARFI imaging, the Von Mises stresses reported herein for the baseline model are compared in Table 3 with the peak Von Mises stress due to blood pressure reported by Li et al. (Li et al., 2006), whose published histology and MRI images were used to define the FEM meshes used in this study. Li et al. (Li et al., 2006) found higher peak stresses in Models A and E, which were classified as ruptured plaques based on evidence from *in vivo* MRI. Similarly, Cheng et al. (Cheng et al., 1993) reported an average peak circumferential stress of $545.3 \pm 159.8 \text{ kPa}$ in ruptured lesions due to arterial pressure, which was significantly higher than $192.5 \pm 64.7 \text{ kPa}$ found in unruptured lesions. Loree et al. (Loree et al., 1992) have argued that fracture of the fibrous cap generally does not occur at circumferential stresses lower than 300 kPa, based on uniaxial elongation tensile tests of fibrous caps from ruptured and unruptured aortic plaque specimens (Lendon et al., 1991). Despite observed variations in the magnitude and location of developed stresses as a function of geometry and material properties, the Von Mises stresses we have modeled in response to an ARFI excitation were $< 1.2 \text{ kPa}$ for all parametric cases and geometries considered.

Disparity between the magnitude of stresses induced by an ARFI excitation compared to blood pressure are likely due to differences that exist in the loading of the different force states. For pressurized thin walled cylinders the circumferential stresses are dominant according to Laplace's Law (Arroyo and Lee, 1999) which may explain why areas of peak concentration in thin fibrous caps due to blood pressure have been linked to sites of rupture (Cheng et al., 1993). Unlike blood pressure, however, ARFI imaging applies a compressive normal force acting in a single radial direction. It was observed herein that areas of increased stress occur at the edges of the applied acoustic radiation force, as in Figure 2c, suggesting that shear stresses are more dominant with ARFI imaging.

4.3 Clinical Utility

The FEM models developed in this study demonstrate feasibility for safely identifying vulnerable plaques with ARFI imaging and allow for a better understanding of how we might expect the method to perform in cases of varying atherosclerotic burden. While it was demonstrated that ARFI imaging can readily differentiate a soft region within an otherwise normally stiff media, it's unlikely that the method can be used distinguish the soft region as an area of intraplaque hemorrhage and/or a lipid pool. Future studies comparing ARFI derived metrics of stiffness to gold standard methods for plaque component characterization, such as histology or MRI, will assess the clinical utility of the method and allow for validation of the FEM simulations performed herein.

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Figure 1.

A FEM simulation portraying the axial displacements from a single, impulsive, acoustic radiation force excitation centered laterally in a 3-D, linear, elastic, homogeneous, and isotropic solid with a Young's modulus of 4 kPa. The response as a function of axial depth and time following excitation is shown in a) where the white-dashed line indicates the time of maximum axial displacement. The maximum axial displacements as a function of lateral and axial position that occur through time is shown in b) for a single excitation. The synthesized 2-D ARFI maximum axial displacement image created by combining the simulated response from nine separate excitations applied at different lateral locations is shown in c).

Figure 2. Simulated FEM results obtained using the baseline material properties ($\alpha = 0.5$ dB/cm/MHz, $E_{media} = 1000$ kPa, $E_{cap} = 1000$ kPa, $E_{lipid} = 25$ kPa) for the five carotid artery models investigated. a) Component segmentation of the media, lumen, lipid, fibrous cap, and surrounding tissue components. b) Maximum axial displacement images allow for the visualization of the lipid pool, which displaces less, compared to the stiffer surrounding media and fibrous cap. c) Maximum Von Mises stress contours show regions of stress concentrations in the thin fibrous cap and media regions.

Figure 3. Simulated FEM results comparing the effects of the attenuation coefficient ($\alpha = 0.3, 0.5$, and 0.7 dB/cm/MHz) on the maximum axial displacements and Von Mises stresses in Model D with material properties $E_{media} = 1000$ kPa, $E_{cap} = 1000$ kPa, and $E_{lipid} = 25$ kPa.

Figure 4.

Simulated FEM results comparing the effects of the media Young's modulus values ($E_{\text{media}} = 200, 600, 1000, 1500, \text{ and } 2000 \text{ kPa}$) on the maximum axial displacements and Von Mises stresses in Model B with material properties $\alpha = 0.5 \text{ dB/cm/MHz}$, $E_{\text{cap}} = 1000 \text{ kPa}$, and $E_{\text{lipid}} = 25 \text{ kPa}$.

Figure 5.

Simulated FEM results comparing the effects of the fibrous cap Young's modulus values ($E_{\text{cap}} = 1000, 2000, 3000, 4000$, and 5000 kPa) on the maximum axial displacements and Von Mises stresses in Model E with material properties $\alpha = 0.5$ dB/cm/MHz, $E_{\text{media}} = 1000$ kPa, and $E_{\text{lipid}} = 25$ kPa.

Figure 6.

Simulated FEM results comparing the effects of the lipid Young's modulus values ($E_{lipid} = 1000, 1250, 1500, 1750, \text{ and } 2000 \text{ kPa}$) on the maximum axial displacements and Von Mises stresses in Model A with material properties $\alpha = 0.5 \text{ dB/cm/MHz}$, $E_{media} = 1000 \text{ kPa}$, and $E_{cap} = 1000 \text{ kPa}$.

Figure 7.

Summary results comparing 1) the contrast of the lipid pool compared to the media and fibrous cap components and 2) the maximum simulated Von Mises stress values in the media and fibrous cap regions for each performed simulation.

Table 1

Assigned Static Young's Modulus Values

	Baseline	Modified
E_{media} (kPa)	1000	200, 600, 1500, 2000
E_{cap} (kPa)	1000	2000, 3000, 4000, 5000
E_{lipid} (kPa)	25	0.5, 1, 5, 50

Table 2

Simulated Ultrasound Transducer Parameters

Frequency (MHz)	4.0
Pulse Duration (# cycles)	600
Transmit F/#	3.0
Focal Depth (mm)	20

Table 3

Comparison of the Maximum Von Mises Stress Induced by ARFI and Blood Pressure

	ARFI Induced Von Mises Stress (Pa)	Blood Pressure Induced Von Mises Stress (Pa)*
Model A	603	694,100
Model B	435	199,600
Model C	663	161,100
Model D	837	319,200
Model E	622	672,500

* Previously reported by Li et al., 2006