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Monitoring Microwave Ablation of *Ex-Vivo* Bovine Liver Using Ultrasonic Attenuation Imaging

Kayvan Samimi¹, James K. White², Christopher L. Brace², and Tomy Varghese^{1,3}

¹Department of Electrical and Computer Engineering, College of Engineering

²Department of Radiology, University of Wisconsin School of Medicine and Public Health

³Department of Medical Physics, University of Wisconsin School of Medicine and Public Health, University of Wisconsin—Madison, 1111 Highland Avenue, Room 1122 WIMR, Madison, Wisconsin 53705

Abstract

Thermal ablation of soft tissue changes the tissue microstructure, and consequently induces changes to its acoustic properties. While B-mode ultrasound provides high-resolution and high-frame-rate images of ablative therapeutic procedures, it is not particularly effective at delineating boundaries of ablated regions due to poor contrast in echogenicity between ablated and surrounding normal tissue. Quantitative Ultrasound (QUS) techniques can provide quantitative estimates of acoustic properties such as backscatter and attenuation coefficients and differentiate ablated and unablated regions more effectively, with the potential for monitoring minimally invasive thermal therapies. In this study, a previously introduced attenuation estimation method was used to create quantitative attenuation coefficient maps for 11 microwave ablation procedures performed on refrigerated *ex-vivo* bovine liver. The attenuation images correlate well with the pathological images of the ablated region. The mean attenuation coefficient for regions of interest (ROIs) drawn inside and outside the ablated zones were 0.9 (± 0.2) and 0.45 (± 0.15) dB/cm/MHz, respectively. These estimates agree with reported values in the literature and establish the usefulness of noninvasive attenuation imaging for monitoring therapeutic procedures in the liver.

Keywords

ultrasonic attenuation coefficient; thermal ablation; microwave ablation; spectral shift

Introduction

Heat-based thermal ablation is a minimally invasive alternative to surgical resection for treating patients with primary and metastatic tumors of the liver (Liang et al. 2005; Liang

Corresponding Author: Kayvan Samimi, 1111 Highland Avenue, Room 1122 WIMR, Madison, WI 53705, USA. samimi@wisc.edu, Phone: +1 (608)-692-9026.

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and Wang 2007; Seki et al. 1994; Shibata et al. 2002). A small needle-like applicator is inserted into the patient's abdomen via a small incision and used to deliver radio-frequency or microwave energy in and around the tumor (Seki et al. 1994). The energy is absorbed into the surrounding tissue and heats a roughly ellipsoidal zone to cytotoxic temperatures (50 °C for a few minutes). The ideal ablation zone would include the tumor and a margin of normal tissue that is analogous to a surgical margin (Ahmed et al. 2014; Nakazawa et al. 2007).

Accurate and reliable imaging is critical throughout all stages of a percutaneous ablation procedure. This includes feedback during applicator insertion, real-time monitoring of ablation growth during tissue heating, and immediate assessment of the ablation zone post procedure (Liang and Wang 2007). Computed tomography (CT) and magnetic resonance imaging (MRI) are effective tools for guidance and follow-up assessment of ablation treatments (de Senneville et al. 2007; Goldberg et al. 2000; Hyodoh et al. 1998; Park et al. 2009; Weidensteiner et al. 2003). Both provide adequate resolution and volumetric information about the ablation zone, but are less practical in real-time assessment of ablation growth. Furthermore, CT exposes the patient and clinicians to ionizing radiation, and MRI hardware incompatibilities require the removal of any applicators prior to imaging, complicating the ability to reposition and re-treat lesions if needed.

Current practice often utilizes conventional B-mode ultrasound imaging (Chiou et al. 2007), as it provides effective real-time feedback during applicator positioning as well as feedback based on a spectrum of mechanical and acoustical changes resulting from tissue heating (Dong et al. 2003). At the onset of ablation, thermal expansion and phase transitions due to chemical and structural protein modifications occur in the heated tissue. Using B-mode ultrasound, this early stage of coagulation and cell death presents as a subtle hypoechoic zone (Cha et al. 2000). Temperatures climb and diffuse outwardly as the ablation persists. Dissolved gases in the tissue release and water vapor evacuates to form a zone of hyperechoic microbubbles, allowing room for the tissue to contract toward the center and stiffen (Brace et al. 2010). The bubbles also give rise to acoustic shadowing below the thermal lesion (Lewis et al. 2015). It has been demonstrated (Bush et al. 1993; Cha et al. 2000; Wells et al. 2015) that the hyperechoic zone has some correlation with the location of ablation, but the shape and size of regions of high backscattering coefficient correspond poorly with the shape and size of the ablated lesions. Additionally, changes in scattering are associated with the transient presence of gas bubbles and not the ablated tissue itself, which makes it difficult to perform any post-treatment assessment based on scattering changes. Therefore, a more precise method of visualizing the ablated zone and its boundary is needed that would allow for more optimal treatment of the tumor mass and margin while minimizing risk to vulnerable areas.

Quantitative Ultrasound (QUS) imaging methods can visualize mechanical and acoustic tissue properties other than echogenicity. As such, they hold great promise for producing high-contrast images for diagnosis and treatment monitoring. In the particular case of thermal ablation of tumors, protein denaturation and changes to tissue microstructure cause an increase in the elastic modulus of tissue in the ablated zone. As a result, several QUS elasticity imaging methods have been investigated for their ability to differentiate the stiffer

ablated core from the softer surrounding tissue. Some of these imaging methods that have shown good correlation with the gross pathology of ablation include strain imaging (Kallel et al. 1999; Rubert et al. 2010; Varghese et al. 2002), sonoelastography (Zhang et al. 2008), acoustic radiation force imaging (Fahey et al. 2006), and elastic modulus imaging (Jiang et al. 2010).

Ultrasonic attenuation coefficient is another QUS parameter of interest that has been studied extensively for its potential diagnostic value. It has been evaluated as a classifier for diffuse liver diseases (Gaitini et al. 2004; Ghoshal et al. 2012; Lin et al. 1988; Narayana and Ophir 1983), myocardial conditions (Mimbs et al. 1977; O'Donnell et al. 1981), breast cancers (Chang et al. 2007), arterial plaques (Bridal et al. 2000; Shi et al. 2008), deep vein thrombosis (Franceschini et al. 2010; Yu and Cloutier 2007), osteoporosis (Wear 1999, 2001), and pathological conditions in other tissue types (Bridal et al. 2006). In addition to its diagnostic value, estimation and compensation for attenuation can lead to improvements in the estimation accuracy of other QUS parameters (Lizzi et al. 2003; Nam et al. 2012; Vlad et al. 2008). Concerning the effects of thermal ablation, many investigators have reported significant increases in the ultrasonic attenuation coefficient of ablated soft tissues.

Converted to the units used in this paper, their reported values for the slope of attenuation coefficient with frequency are as follows. Parker (1983) used a variety of measurement techniques and found that the attenuation coefficient for refrigerated bovine liver has a mean value of 0.4 dB/cm/MHz with a sample-to-sample standard deviation of 0.09 dB/cm/MHz in the typical ultrasonic frequency range of 1 to 6 MHz. Damianou et al. (1997) studied canine soft tissues at temperatures ranging from room temperature to 70 °C and found that the attenuation coefficient of degassed *ex-vivo* canine liver drops from 0.4 dB/cm/MHz at room temperature to a minimum of 0.3 dB/cm/MHz at 40 °C and then sharply increases at ablative temperatures of 50 °C and above to a maximum of 0.73 dB/cm/MHz at 70 °C. Damianou et al. did not study higher temperatures to avoid formation of gas bubbles. The inter-sample variation in their measurements was around 10%. Gertner et al. (1997) studied the effects of heating from 37 °C to 70 °C on refrigerated and degassed bovine liver in the frequency range of 2.5 to 5 MHz. Interpreting their results in terms of slope of attenuation with frequency, a measurement of 0.7 dB/cm/MHz for unheated liver (37 °C) and 0.9 dB/cm/MHz for heated liver (70 °C) is obtained. However, based on the standard errors of the reported frequency-dependent attenuation estimates, these slopes could have a standard error as high as 30%. Yet, it is notable that Gertner et al. also observed a decrease in attenuation (at 3.5 MHz) by about 20% while the liver was warmed to sub-ablative temperatures (< 50 °C), followed by an 85% increase in attenuation as the liver was further heated to 70 °C. Bamber and Hill (1979) reported similar findings for freshly excised and degassed bovine liver. Techavipoo et al. (2004) studied freshly excised canine liver samples. They also reported a comparable trend of attenuation decrease as the liver samples were warmed to sub-ablative temperatures, followed by attenuation increase in ablative temperatures. However, the standard errors of their reported values were too high to allow for interpretation of slope of attenuation with frequency.

In this paper, instead of measuring bulk attenuation changes using a through-transmission insertion loss approach, we focus on producing local attenuation maps and images from backscattered radiofrequency (RF) data acquired using clinical linear-array ultrasound

scanners. To this end, a previously introduced spectral-shift attenuation estimation method (Samimi and Varghese 2016b) was applied to RF data that were collected during microwave ablation of bovine liver samples. The resulting attenuation images are compared to gross pathology. Finally, numerical values of these attenuation coefficient estimates in ablated and unablated regions of interest (ROIs) are compared to the reported values from the literature.

Materials and Methods

Experimental Setup and Data Collection

Microwave ablation procedures were performed on 11 samples of refrigerated bovine liver tissue. Tissue samples were acquired from 5 separate whole livers that were bought from a local slaughterhouse. Experiments were conducted within 4 days of liver excisions. The samples were roughly cubic with approximate dimensions of $6 \times 6 \times 6$ cm. A sample holder platform with two metal rods was used to fix the liver samples in place. For each ablation procedure, the microwave ablation probe (i.e., microwave antenna) was inserted into the center of the liver sample under the guidance of B-mode ultrasound imaging. A Certus 140 microwave ablation system (NeuWave Medical; Madison, WI, USA) equipped with a Certus PR ablation probe was used to ablate liver samples and an Acuson S2000 ultrasound system (Siemens Medical Solutions USA Inc.; Mountain View, CA, USA) equipped with a 9L4 linear array transducer was used to collect B-mode and RF ultrasound data. Once the ablation probe was in place, the ultrasound transducer was fixed on top of the liver surface using a flexible arm clamp, such that the imaging plane captured the tip of the ablation probe as it entered the plane at a small angle with the transducer axis. Four (4) fiber optic temperature probes (Neoptix; Quebec City, QC, Canada) were inserted into the imaging plane perpendicularly, in order to record the temperature of the ablation zone in real time. Ablation duration was set to 4 minutes and microwave output power was set to 55 Watts. Frames of RF data were collected at 2 second intervals for the duration of ablation, and then at 5 second intervals for 26 minutes of cool down for a total of 30 minutes. Simultaneously, temperature readings from the fiber optic probes were recorded every 1 second. At the end of the procedure, a well-characterized tissue-mimicking (TM) reference phantom with uniform speed of sound (1540 m/s), scattering, and attenuation (0.5 dB/cm/MHz) properties was scanned using the same scanner settings that were used for the liver (Yao et al. 1990). In these experiments, the 9L4 transducer was operated in penetration mode (with 100% transmit power) and with a transmit center frequency of 6 MHz and a -3 dB bandwidth of 50%. Finally, the ablated liver samples were cut along the ultrasonic imaging plane and pathological images of the ablated zone were acquired.

Estimation of Attenuation Coefficient from RF Data

RF data were downloaded from the ultrasound scanner and loaded into MATLAB (MathWorks, Inc.; Natick, MA) on a computer workstation for post-processing. A previously introduced frequency-domain attenuation estimation method, termed optimal power-spectral shift estimator (OPSSE), was utilized to estimate local attenuation coefficient and create attenuation images (Samimi and Varghese 2016b). Due to the frequency-dependent nature of attenuation in soft tissues, the spectral content of the transmitted ultrasonic pulse gets attenuated more at the higher frequency end of its spectrum than the

lower end. The net effect of this phenomenon is an apparent shift of the spectral representation of the ultrasonic pulse towards lower frequencies as it propagates through the attenuating medium. The OPSSE is an optimized correlation-based method that estimates this spectral shift with depth, and mathematically translates it to an estimate of the ultrasonic attenuation coefficient. The step-by-step algorithm for implementing this method is given below.

1. Parse sample and reference RF data frames into blocks for spectral analysis. Each block contains multiple data segments from adjacent A-lines.
2. Estimate the average power spectral density for each data block using Thomson's multitaper method (Thomson 1982). These tapers are the discrete prolate spheroidal sequences (DPSS), also known as Slepian sequences.
3. Normalize the power spectra by dividing the sample spectrum from a data block by the reference spectrum computed from the corresponding data block of the reference TM phantom.
4. Identify the usable bandwidth of the normalized spectra. This is the frequency range where both sample and reference spectra are above the noise floor.
5. Multiply the normalized power spectra by the transmit pulse spectrum. If the exact transmit spectrum is unknown, use a Gaussian profile centered at the transmit frequency with a bandwidth similar to the transmit spectrum.
6. Cross-correlate the normalized and filtered power spectra with the reciprocal of the transmit pulse spectrum within the usable bandwidth and find the frequency shift that minimizes the cross-correlation.
7. Perform a linear regression on the estimated frequency shifts from several axially consecutive data blocks within an ROI and estimate the slope of attenuation with depth and frequency (i.e., the attenuation coefficient) for that ROI.

Attenuation maps were generated for all RF data frames using the described algorithm and the following processing parameter settings: data block length = 8 mm, data block width = 36 A-lines, axial data block overlap = 30%, number of data blocks per attenuation estimation ROI = 3, and standard deviation of the Gaussian transmit spectrum = 1 MHz. The choice of signal processing settings for spectral estimation involves a trade-off between the accuracy and stability of the estimated power spectra on one hand, and the spatial resolution of the resulting attenuation image on the other. In this study, the settings were selected such that maximum spatial resolution would be preserved while maintaining stable spectral estimation. In particular, 3-point axial regression and a small gating window (sized 8 mm, i.e., 10 μ s or $\sim$ 6 pulse lengths for this transmit pulse,) ensure high axial resolution, and 36 A-lines per data block is close to the minimum number of spectral averages needed for a stable power spectral estimate and ensures high lateral resolution. A 30% axial overlap between data blocks further boosts axial spatial resolution while minimally impacting the accuracy of the estimates. More technical details regarding this estimation method and the impact of processing parameter settings on estimation accuracy and precision is presented in (Samimi and Varghese 2016b). In order to create displayable attenuation images, the

estimated attenuation maps were interpolated to the same resolution as the B-mode images. Next, they were color-coded and overlaid on the corresponding grayscale B-mode images.

Results

Attenuation images were compared to B-mode and gross pathological images for the 11 ablation procedures. A visual analysis of two sample cases is presented, followed by statistical analysis of the quantitative attenuation coefficient estimates. Statistical analysis of numerical attenuation coefficient estimates was performed on ablated and unablated ROIs drawn inside and outside the ablation zone.

Visual Analysis of Attenuation Images

B-mode and attenuation images for the two sample cases are presented in Figure 1 and Figure 2 at eight (8) different time points during the ablation procedure and post-ablation cool-down. Note the hyperechoic cloud of gas bubbles that forms at the ablation site and grows rapidly as the procedure progresses. The cloud of gas bubbles acts as a congregation of scatterers, increasing both the scatterer number density and also the frequency dependence of scattering based on the bubble dimensions. Due to the diverse range of bubble diameters, different types of scattering and reflection are present. The hyperechogenicity of the gas cloud is caused by the increased number of scatterers. At the same time, multiple reflections cause a bright streak artifact below the gas cloud on the B-mode images. The echo signal backscattered from these bubbles saturates and masks any tissue signal originating from their location or below it. During the cool-down period, some of the bubbles dissipate by outgassing or dissolving back into the liver fluids and more of the backscattered tissue signals reach the transducer. Note that the more strongly attenuating ablated zone begins to cast a shadow below the ablated region as more bubbles dissipate. Observe that the attenuation images successfully track the progression of the ablation procedure. The general absence of negative attenuation estimates is a notable feature of the new OPSSE estimation method when compared to previous estimators (Samimi and Varghese 2016b). However, some underestimation and negative estimates are seen below the highly attenuating ablated zone where the signal to noise ratio (SNR) drops and cross-correlation analysis fails to locate the true spectral centroid. Comparing B-mode and attenuation images in Figure 1 at the 30-minute mark reveals that this artifact is only present where the available tissue signal is very weak. In such areas, the spectral estimator is effectively operating on noise.

Based on the microwave radiation pattern of the applicator used in this study, the expected shape of the ablated zone is an ellipsoid with its long axis along the microwave ablation probe. In order to evaluate the usefulness of the attenuation images for monitoring ablation procedures, a multi-observer study was performed. While blinded to gross pathological images, all authors segmented the ablated zone on post-ablation attenuation images by visually fitting an ellipse to the high-attenuation area of the attenuation map at the end of the 30-minute procedure.

Observers were instructed to include higher-than-background attenuation coefficient estimates (warmer colors) as well as a potential artifact-induced low-attenuation area (cooler

colors) directly below it in their segmented lesion. Similarly, observers were asked to fit an ellipse to the hyperechoic cloud of bubbles on the B-mode images. The images were de-identified and their order was randomized for this task. The area and the lateral diameter of these ellipsoids were compared to the pathological image of the ablated zone using the image processing program ImageJ (Rasband, W.S., U. S. National Institutes of Health, Bethesda, MD, USA). Examples of these comparisons are presented in Figure 1c and Figure 2c for the two sample cases. It appears that the bubble cloud does not necessarily provide an accurate estimate of the lateral extent of the ablation zone or its boundaries. The attenuation image appears to have a better agreement with the pathological image. This process was repeated for all 11 cases in order to numerically compare the correlation between pathological images (i.e., ground truth) and attenuation or B-mode images in terms of their associated estimated area and lateral diameter of the ablation zone. Figures 3 and 4 present the regression analyses for the ablated area and the lateral extent of ablation, respectively. In these plots, the solid black line with unit slope represents a perfect correlation between averaged observer measurements from imaging and gross pathology. The error bars show the standard deviation of observer measurements for each ablation experiment. Observe that, in the area plot (Figure 3), measurements from attenuation images show good correlation with measurements from gross pathology ($r = 0.76$, $p < 0.006$). On the other hand, measurements from B-mode images do not show any statistically significant correlation with pathological measurements of ablated area. In the lateral diameter plot (Figure 4), measurements from attenuation images show a weak correlation with pathological measurements ($r = 0.46$, $p < 0.15$) while measurements from B-mode images do not show any statistically significant correlation with pathology. It is notable that observer measurements on attenuation images have smaller standard deviations when compared to measurements on B-mode images.

Figures 5 and 6 illustrate the placement of fiber optic temperature probes on the pre-ablation B-mode image, as well as the post-ablation B-mode image, and the plot showing the temperature readings. Notice that in Figure 5 temperature probes 1 and 4 are placed farther from the microwave ablation probe and do not reach cytotoxic temperatures of 50 °C or more. Therefore, their locations are outside the ablated zone. However, an observer looking at the cloud of bubbles on the B-mode image alone might place them inside the ablated zone.

Statistical Analysis of Attenuation Estimates

To perform a statistical analysis of the attenuation coefficient estimates, two sets of ROIs were drawn on the un-interpolated post-ablation attenuation map. Each ROI contained a minimum of 15 independent attenuation coefficient estimates in order to provide significant statistical estimates. Ablated ROIs were drawn such that they would include the top half of the ablated zone and exclude the low-SNR bottom half of the ablated zone. Similarly-sized unablated ROIs were drawn outside the ablated zone. Examples of these ROIs are presented in Figure 7a for the first sample case and in Figure 8a for the second sample case. Statistics of the estimated attenuation coefficient in these ROIs (i.e., mean and standard deviation) were calculated for the 30-minute time course of each procedure. Figures 7b and 8b show the mean attenuation coefficient plots with 99% confidence intervals (± 1 standard deviation) for the first and second sample cases, respectively. Note that as the liver sample cools down

and the gas bubbles dissipate, the attenuation estimates for the ablated ROI become more stable and approach their true value.

Statistically significant values for the attenuation coefficient of ablated and unablated *ex-vivo* bovine liver were obtained after synchronizing the estimates from all 11 procedures and calculating the mean and the standard deviation of the attenuation coefficients in ablated and unablated ROIs at various time points. Synchronization was performed by converting frame numbers to time stamps with Time = 0 marking the start of heating. These results are presented in Figure 9. The final attenuation coefficient estimates of 0.9 ± 0.2 dB/cm/MHz for ablated liver and 0.45 ± 0.15 dB/cm/MHz for unablated liver are comparable to reported values in the literature that were discussed in the introduction section (Damianou et al. 1997; Gertner et al. 1997; Parker 1983; Techavipoo et al. 2004). These results are significant as they present a meaningful separation of the attenuation coefficient values between ablated and unablated liver. Thus, they can be used to classify liver tissue undergoing ablation treatment as either ablated lesion or unablated background tissue.

Discussion

B-mode images do not provide significant contrast between ablated and unablated liver due to their similar echogenicity. Gertner et al. (1997) reported that changes in the backscatter coefficient as a result of heating are small and of the same order as the natural variations of this parameter from point to point in the liver tissue. On the other hand, the gas bubbles that form at ablative temperatures are highly echogenic. However, their location and distribution within the imaging plane is a poor indicator of the ablated region and its boundary, since these bubbles are not confined to the ablated region and actively escape the high-pressure center of the ablation zone, often with an audible popping sound.

Compared to B-mode images, attenuation images demonstrate a better correlation with gross pathological images. Many authors have reported on an irreversible increase in the ultrasonic attenuation (by about 100%) as a result of heating and protein denaturation in soft tissues (Damianou et al. 1997; Gertner et al. 1997). The results of this paper corroborate their findings. Figure 9 shows a slight decrease of the attenuation coefficient as the liver samples warm up from refrigerated temperatures to sub-ablative temperatures, followed by a rapid increase of the attenuation coefficient in the ablated ROIs as heating continues. Bamber et al. (1977) cautioned that formation of gas bubbles can lead to increased attenuation and inflate the estimates by up to a factor of 3. Elevated attenuation estimates corresponding to the cloud of gas bubbles are also noted in our attenuation images during the heating period. However, as the liver sample cools down and the gas bubbles begin to dissipate, the attenuation image stabilizes and begins to reflect the underlying tissue attenuation. It should be noted that most of the reported attenuation measurements in the literature have been obtained using the substitution (or insertion loss) technique by comparing the spectral amplitude of the reflected signal with and without the liver sample in the acoustic path. The estimation method of this paper does not rely on the spectral amplitude; rather, it estimates the attenuation coefficient from the overall spectral shape and its centroid. Therefore, this method is less susceptible to amplitude changes that result from the formation of gas

bubbles. Other spectral-based attenuation estimation methods are discussed and analyzed in previous papers (Samimi and Varghese 2015, 2016a).

We found that attenuation estimates continue to climb during the cool-down period following the completion of heating. This observation can be attributed to dissipation of gas bubbles after ablation. Due to nonlinear scattering of ultrasound by microbubbles, some of the backscattered ultrasound energy is transferred to harmonic frequencies (Miller 1981). This phenomenon could affect the estimated spectral centroids and bias the attenuation estimates. Additionally, the backscattered signal from the gas bubbles dominates the tissue signal during the heating period.

Therefore, it was concluded that reliable attenuation images representative of the actual tissue state (i.e., ablated vs. unablated) can only be obtained after a cool-down period that ensures dissipation of gas bubbles. This cool-down period may be shorter for *in-vivo* ablations where blood circulation provides an active heat sink.

Conclusion

Common clinical practice in ablative treatment of tumors is to ablate the tumor along with a normal tissue margin of at least 5 mm or more in order to minimize risk of local tumor recurrence (Wang et al. 2013). B-mode imaging tends to overestimate the ablated area and boundaries of the ablated region. Therefore, it cannot verify with high certainty that the tumor and a normal tissue margin have been successfully ablated. In this study, local attenuation imaging using an optimized spectral shift estimation method was used to monitor microwave ablation in *ex-vivo* bovine liver. Attenuation images were shown to correlate with gross pathology of the ablation better than conventional B-mode images. Thus, establishing the usefulness of attenuation imaging for clinical monitoring of minimally invasive thermal therapies.

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Figure 1. Images for the first sample case: (a) B-mode ultrasound images of ablation at 8 time points; (b) Attenuation images of ablation at 8 time points; (c) Comparison between post-procedure attenuation image, B-mode image, and gross pathological image.

Figure 2.

Images for the second sample case: (a) B-mode ultrasound images of ablation at 8 time points; (b) Attenuation images of ablation at 8 time points; (c) Comparison between post-procedure attenuation image, B-mode image, and gross pathological image.

Figure 3.

Comparison of ablated area measured from gross pathology with ablated area measured from attenuation images (circles) or B-mode images (squares). Error bars show the standard deviation of observer measurements. Area measurements from attenuation images have a good correlation with pathology ($n = 11$, $r = 0.76$, $p < 0.006$). Solid line with unit slope represents perfect agreement. Measurements from B-mode images show no statistically significant correlation with pathology.

Figure 4.

Comparison of lateral diameter of ablation measured from gross pathology with lateral diameter of ablation measured from attenuation images (circles) or B-mode images (squares). Error bars show the standard deviation of observer measurements. Diameter measurements from attenuation images have a weak correlation with pathology ($n = 11$, $r = 0.46$, $p < 0.15$). Solid line with unit slope represents perfect agreement. Measurements from B-mode images show no statistically significant correlation with pathology.

Figure 5. Fiber optic monitoring of temperature changes for the first sample case: (a) Pre-ablation placement of ablation antenna and fiber optic temperature probes; (b) Post-ablation B-mode image; (c) Plot of temperature measurements for the 30-minute course of the procedure.

Figure 6.

Fiber optic monitoring of temperature changes for the second sample case: (a) Pre-ablation placement of ablation antenna and fiber optic temperature probes; (b) Post-ablation B-mode image; (c) Plot of temperature measurements for the 30-minute course of the procedure.

Figure 7.

(a) Placement of ablated and unablated regions of interest (ROIs) with respect to the un-interpolated attenuation map of the first sample case. (b) Plot of mean attenuation coefficient in the two ROIs over the 30-minute course of the procedure along with 99% confidence intervals of ± 1 standard deviation. 4 minutes of active heating ends at frame 137.

Figure 8.

(a) Placement of ablated and unablated regions of interest (ROIs) with respect to the un-interpolated attenuation map of the second sample case. (b) Plot of mean attenuation coefficient in the two ROIs over the 30-minute course of the procedure along with 99% confidence intervals of ± 1 standard deviation. 4 minutes of active heating ends at frame 42.

Figure 9.

Mean attenuation coefficient of $n = 11$ ablated regions of interest (ROIs) (diamonds) and $n = 11$ unablated ROIs (squares) over the 30-minute course of the procedures along with error bars representing one standard deviation. Reportable values of $0.9 (\pm 0.2)$ dB/cm/MHz for ablated and $0.45 (\pm 0.15)$ dB/cm/MHz for unablated refrigerated bovine liver are obtained.