***In vivo* evaluation of a breast-specific magnetic resonance guided focused ultrasound system in a goat udder model**

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Purpose: This work further evaluates the functionality, efficacy, and safety of a new breast-specific magnetic resonance guided high intensity focused ultrasound (MRgFUS) system in an *in vivo* goat udder model.

Methods: Eight female goats underwent an MRgFUS ablation procedure using the breast-specific MRgFUS system. Tissue classification was achieved through the 3D magnetic resonance imaging (MRI) acquisition of several contrasts (T1w, T2w, PDw, 3-point Dixon). The MRgFUS treatment was performed with a grid trajectory executed in one or two planes within the glandular tissue of the goat udder. Temperature was monitored using a 3D proton resonance frequency (PRF) MRI technique. Delayed contrast enhanced-MR images were acquired immediately and 14 days post MRgFUS treatment. A localized tissue excision was performed in one animal and histological analysis was performed. Animals were available for adoption at the conclusion of the study.

Results: The breast-specific MRgFUS system was able to ablate regions ranging in size from 0.4 to 3.6 cm³ in the goat udder model. Tissue damage was confirmed through the correlation of thermal dose measurements obtained with realtime 3D MR thermometry to delayed contrast enhanced-MR images immediately after the treatment and 14 days postablation. In general, lesions were longer in the ultrasound propagation direction, which is consistent with the dimensions of the ultrasound focal spot. Thermal dose volumes had better agreement with nonenhancing areas of the DCE-MRI images obtained 14 days after the MRgFUS treatment.

Conclusions: The system was able to successfully ablate lesions up to 3.6 cm³. The thermal dose volume was found to correlate better with the 14-day postablation nonenhancing delayed contrast enhanced-MR image volumes. While the goat udder is not an ideal model for the human breast, this study has proven the feasibility of using this system on a wide variety of udder shapes and sizes, demonstrating the flexibility that would be required in order to treat human subjects. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4811103>]

Key words: focused ultrasound, breast cancer, magnetic resonance, thermometry

I. INTRODUCTION

Breast cancer remains an important disease, adversely affecting a large population of women. The chance of developing breast cancer at some time in a woman's life is approximately 1 in 8.¹ Despite more sensitive magnetic resonance imaging (MRI) methods to detect breast lesions² and new min-

imally invasive conservative forms of therapy³⁻⁸ there still remains significant room for improvement in both imaging and therapy. MRI-guided focused ultrasound (MRgFUS) has the potential to provide completely noninvasive therapy of localized breast cancer, avoiding the discomfort and potential complications associated with general anesthesia and surgery. MRgFUS has the potential to improve cosmetic outcomes by

avoiding scarring, and possibly allow earlier administration of systemic therapies due to shorter postoperative recovery time. Indeed, a minimally invasive treatment such as MRgFUS may be the only viable treatment recourse when radiation or chemotherapy limits have been exhausted.

A unique capability of MRI is the ability to monitor and assess thermal therapy treatments in near realtime. The proton resonance frequency (PRF) technique is currently the most consistent and efficient method for measuring temperature changes in water-based tissues.^{9–13} Spatially varying thermal dose values¹⁴ calculated from dynamic MRI temperature measurements are used as an indicator to determine tissue viability during the ablation procedure. T2-weighted and contrast enhanced T1-weighted images have been used to assess tissue viability posttreatment.¹⁵ Some studies have found that the nonenhancing volume measured after an ablation treatment is larger than what is predicted by thermal dose measurements. While better agreement is generally seen several days after treatment,¹⁶ discrepancies are still present.^{17–19} Several potential reasons are noted for these discrepancies. The occlusion of vessels may cause ischemic necrosis in tissues not directly ablated. The underestimation of temperature measurements would also lead to a smaller volume reaching thermal dose ablation levels.¹⁷ Lenard *et al.*¹⁹ observed that hypointensity in uterine fibroids indicated a low cellularity and vascularity making the tissue more susceptible to focused ultrasound ablation while fibroids exhibiting hyperintensity had increased vascularity making them more difficult to ablate.

Several preliminary MRgFUS studies have shown the feasibility of using this technology to treat localized breast lesions.^{20–24} Most studies have used a vertically propagating transducer, which can limit the areas that can be treated effectively with MRgFUS. There have been studies that have demonstrated that the use of a laterally-shooting transducer could potentially treat a larger variety of locations in the breast.^{21,25–28} Recently, a new breast-specific magnetic resonance guided high intensity focused ultrasound system has been designed and evaluated in phantoms and *ex vivo* tissues.²⁹ This system has several unique features including a transducer that is laterally mounted onto a treatment cylinder, an integrated 11-channel RF coil,³⁰ and tensioning device. This work further evaluates the functionality, efficacy and safety of this system in an *in vivo* goat udder model.

II. METHODS

II.A. Animal model

Eight female goats, both lactating and nonlactating (22–48 kg), were used in this study. The Institutional Animal Care and Use Committee approved all experiments. The goal of this study was to treat udders of varied sizes and shapes in order to adequately model the diverse breast sizes that exist in the human population. Early on in the study, it was decided to include lactating goats since those animals had a significantly larger udder size than nonlactating animals. The mean

udder width and length, measured using MRI images were 10.6 (range 7.6–13.3) and 6.8 (range 3.1–12.4) cm. The udder width was measured at the abdominal wall and the length was measured at the udder midline. While udder size was the main factor in determining whether to include an animal in the study, abdominal girth was also a determining factor due to the space limitations of the MRI bore. All pregnant animals were excluded.

Animals were fasted a minimum of 12 h before the administration of anesthesia and were given oxytetracycline (500 mg) in order to suppress the rumen bacteria and minimize rumen production and subsequent bloating during the procedure. Anesthesia was induced using a combination of Ketamine (0.6 mg/kg, IV) and Diazepam (0.2 mg/kg, IV). After intubation, the level of anesthesia was maintained with isoflurane (0.5%–5%, inhaled). All hair was removed from the udder using both clippers and a depilatory cream (Nair, Church and Dwight Co.). An oro-rumen tube was inserted in the esophagus to facilitate rumen drainage. The animal was actively monitored while under anesthesia (body temperature, end tidal CO₂, SpO₂, blood pressure). Postprocedural pain was managed by administering Ketorolac (1.0 mg/kg, SC) preemptively before the procedure. Animals were given three days of additional Ketorolac or Buprenorphine (0.05–0.1 mg/kg, SC) based on the attending veterinarian's recommendations.

II.B. Treatment system

A breast-specific MRgFUS system developed in collaboration with the University of Utah and Image Guided Therapy, Inc. was used for all experiments.²⁹ The system has a laterally mounted 256-element phased-array transducer (Imasonics, Bensacon, France, 1 MHz, 10 cm radius of curvature), a custom 11-channel integrated RF-coil³⁰ and accompanying patient table. The transducer can be moved mechanically with three degrees-of-freedom after the subject is positioned on the table. The system, shown in Fig. 1 is designed to sit atop an existing MRI exam table. All experiments were conducted in a Siemens 3T TIM Trio MRI scanner 60 cm bore size). The distance between the table top and top of the bore was 28 cm. Because of this limitation, animals with a maximum abdominal circumference of larger than 80 cm were not eligible for the study.

II.C. Experimental protocol

The animal was placed on the system in a feet-first prone position with the udder suspended in the treatment cylinder. Before the MRgFUS treatment, several contrasts (T1w, T2w, PDw, 3-point Dixon) of 3D anatomical images were acquired using parameters defined in Wang *et al.*³¹ This data was used to plan the treatment and to segment and classify the different tissue types during postprocessing after the treatment completion. The ultrasound transducer was oriented such that the acoustic window was located at the rear of the udder. Targeted ablation regions were chosen to be in mammary gland tissue at least 1 cm away from both the skin and abdominal

FIG. 1. Breast MRgFUS system used in all experiments placed on the table of a Siemens 3T Trio MRI scanner. The animal was placed in a rear-first prone position.

wall. Mechanical transducer position adjustments were made so that the geometric focus of the transducer was located at the center of the chosen target ablation area. The experimental setup, showing the position of the transducer and orientation of the MR thermometry acquisition is shown in Fig. 2.

The MRgFUS focal spot was initially localized in the animal by applying a low level heating pulse (~ 17 acoustic W, 20–30 s) and observing the resulting temperature rise on MR thermometry images. If positioning adjustments were deemed necessary, the transducer was mechanically moved, or the focal spot electronically steered, and the focal spot localization procedure was repeated. The predefined trajectory was executed around this central point. Trajectories were planned to obtain an ablated region size of 1 to 3 cm³. The ablation trajectory was a grid pattern, executed in one to two planes perpendicular to the ultrasound beam axis, with 9 to 25 points per plane. An example of a 16-point trajectory is shown in Fig. 3. All points were spaced 2 mm apart inplane with 10 mm spacing between planes. Transducer power ranged from 24 to 70 acoustic W applied for 30 to 60 s per point. Sonications were executed consecutively in groups with 2–4 min cooling periods applied between groups. The ultrasound parameters for the ablation were chosen based on the temperature response

observed during the localization procedure with the goal to reach a temperature of approximately 60 °C. The ablation procedure, including the cooling period, was monitored in real-time using MR thermometry (3D segmented-EPI, TR/TE = 35/10 ms, FA = 15°, 752 Hz/pixel, 12 slices plus 16.7% slice oversampling, fat saturation, EPI factor = 9, 4.9 s, 2 × 2 × 2 mm resolution, 128 × 90 matrix). Data was zero-filled interpolated to 1 mm isotropic spacing and reconstructed using the reference-less technique (5th order polynomial, 2D implementation).³² Adjustments to the treatment parameters were made by the operator based on the measured peak temperature rise. Typically changes were made if the peak temperature was greater than 70 °C or less than 55 °C. The treatment specifics for all animals are shown in Table I.

After the ablation procedure, gadolinium contrast was administered (0.1 mmol/kg, Multihance, Bracco Diagnostics) and three-dimensional T1-weighted images (3D VIBE, TR/TE = 11.2/2.63 ms, 12° flip angle, 192 × 192 × 120 matrix, 1 × 1 × 1 mm, 208 Hz/pixel) were acquired to determine the volume of nonenhancing tissue present. This postablation imaging procedure was repeated 14 days after the ablation treatment. The volume of nonenhancing tissue was calculated through segmentation performed manually by 2 to 3 independent observers. The reported value represents an average of those estimates. One goat (animal 2) had the ablated tissue surgically removed 20 days after the ablation procedure and histopathological analysis was performed. All goats were available for adoption at the completion of the study.

FIG. 2. Goat udder suspended in the treatment cylinder. The transducer, seen in the bottom right corner, is tilted up 30° from the horizontal. The MR thermometry slice orientation (solid white box) is parallel to the axis of ultrasound beam propagation.

FIG. 3. Sixteen point, single plane ablation treatment trajectory. (a) Oblique coronal through a goat udder showing the placement of the sonication points. The coronal slice is oriented along the ultrasound beam propagation path as seen in the inset sagittal image. In this example, only one plane is treated, but in one animal (No. 5), two planes were treated, spaced 10 mm apart. (b) Sixteen-point grid trajectory in the transverse plane. Sonications were grouped into four rows. All sonication points were spaced 2 mm apart. Transducer coordinates are indicated in both (a) and (b).

TABLE I. Treatment details for all animals in the study. In animal 5, the multiplane trajectory was executed back-to-front. In animal 4, the multiplane trajectory was executed right-to-left.

Animal No.	Weight (kg)	Lactating (yes/no)	No. of planes	No. of points/plane	Total energy (kJ)
1	48	No	1	9 (3 × 3)	12.15
2	40	No	1	15 (3 × 5)	152.25
3	45	Yes	1	12 (3 × 4)	54.0
4	35	Yes	2	10 (2 × 5), 15 (3 × 5)	225.0
5	22	Yes	2	16 (4 × 4), 16 (4 × 4)	216.0
6	30	Yes	1	20 (5 × 4)	157.5
7	31	Yes	1	16 (4 × 4)	56.4
8	42	Yes	1	9 (3 × 3)	43.875

III. RESULTS

All animals recovered swiftly from the ablation procedure. All lactating animals allowed their kid’s to nurse within 6 h after the procedure. One animal (animal 1) had to be euthanized due to health problems unrelated to the MRgFUS procedure. There were no adverse events related to the MRgFUS procedure with any of the animals and there were no skin burns evident after the ablation procedure.

It was found that nonlactating goats had fat, mammary tissue and milk present in the udder with the fat located at the abdominal wall and between the two mammary glands. Lactating goats had minimal fat between the mammary glands. Both types of animals had milk present in the udder, but goat milk did not appear to negatively affect the beam focus quality. An example of the tissue distribution obtained

FIG. 4. Tissue classification of animal number 1. (a) T2-, T1-weighted and water- and fat-only images from a three-point Dixon technique. (b) Tissue segmentation of air, water, skin, fat, milk, and mammary tissue.

from the first animal is seen in Fig. 4. Using the multiple-contrast images obtained at the beginning of each treatment, the udder tissue was segmented using a hierarchal support vector machine algorithm.³¹

While automatic feedback control was not used during the ablation procedure, realtime temperatures were obtained and actively monitored by the ultrasound operator during the procedure. An example of the temperature response is shown for animal 7 in Fig. 5. The ablation trajectory used with this animal was a 4 × 4 grid pattern, shown in Fig. 3. In this animal, the sonications were applied in four groups, consisting of four adjacent sonication points (Fig. 3). Each sonication group was applied with no cooling period between individual points. Between groups, a cooling period of approximately 4 min was implemented. Figure 5 shows the maximum temperature projections in time for the four sonication groups. Absolute temperature was calculated by adding the animal’s rectal body temperature (39.3 °C) at the start of the ablation procedure to the measured temperature rise. The peak temperature rise achieved varied significantly for each sonication group (65.3 to 85.3 °C). The total energy applied for each group ranged

FIG. 5. Temperature response achieved in animal 7 (4 × 4 trajectory grid in one plane). The maximum temperature response in time for each sonication group is shown in both the oblique coronal and sagittal orientations. The total energy deposited for each group is shown on the respective image. The transducer and MRTI slice orientation is shown in Fig. 2.

FIG. 6. The peak temperature achieved for each sonication group ablated in all animals as a function of total deposited energy for the sonication group.

from 8.4 to 24 kJ, with a total energy deposition of 56.4 kJ. This variability in peak temperature rise was typical of that observed in all animals in the study. Figure 6 shows the peak temperature achieved during all sonication groups as a function of energy deposited for all animals in the study. While there is a general upward trend in peak temperature achieved with increased energy input, there is significant intra and interanimal variability in maximum temperature for each sonication in all animals.

The resulting accumulated thermal dose for animal 7 is shown in Fig. 7. The 3D distribution of the thermal dose is shown in an oblique coronal, axial, and sagittal view. The volume of the lesions, measured as the voxels that achieved a thermal dose of 240 CEM43°C (Ref. 33) or greater was 3.64 cm³. The locations of the sonications are indicated in Fig. 7.

The delayed contrast enhanced images (DCE-MRI) for animal 7 obtained 20 min and 14 days postablation are shown

FIG. 7. Thermal dose accumulated during ablation procedure in animal number 7. The oblique coronal, sagittal and axial views are all shown. Locations of individual sonications are indicated by the white ovals. The total volume that achieved greater than 240 CEM43°C was 3.64 cm³. The transducer location is indicated in the oblique coronal view.

FIG. 8. Delayed contrast enhanced images from animal number 7. (a) 20 min postablation. The nonenhancing volume is 10.13 cm³. Significant edema can be seen around the treatment region. (b) Fourteen-days postablation. The nonenhancing volume is 3.24 cm³ with minimal edema seen.

in Fig. 8. The manually segmented nonenhancing volume 20 min postablation was 10.13 cm³, nearly three times the size of the thermal dose lesion. Fourteen days postablation, the nonenhancing lesion size had reduced to 3.24 cm³. Examples of lesion shapes and sizes ablated in other animals are shown in Fig. 9. The thermal dose accumulated volumes and accompanying DCE-MR images for animals 2, 4, and 6 are shown. Note that two separate lesions were created in animal 4. Thermal dose predicted lesion volumes were 1.23, 2.04, and 3.60 cm³, respectively, compared to nonenhancing volumes on the DCE images of 1.10, 6.27, and 8.21 cm³.

The comparison of the volume that achieved a thermal dose of greater than 240 CEM43°C compared to the 20 min and 14 day postablation nonenhancing volumes for all animals is shown in Fig. 10 in both scatter and Bland-Altman plots. At 20 min post-treatment, for the larger lesions, there is a large variation between the two volumes, with the disparity increasing as the thermal dose calculated volume increases. This variation is reduced in the 14-day postablation data, with the ratio of the two measurements approaching unity across the range of ablated volumes.

In one animal (animal 2), the ablated tissue was excised 20 days after the ablation treatment. Figure 11 shows a clear transition region between normal breast ducts and lobules and adjacent granulation tissue. While a precise registration is not possible, this image shows the side of an ablated region, where the transducer would be located at the bottom left corner of the image. There were also other features of ischemic necrosis including vessel wall thickening, atrophy and hyalinization of tissue.

IV. DISCUSSION

This paper has shown that the breast-specific MRgFUS system is able to ablate regions ranging in size from 0.4 to 3.6 cm³ in an *in vivo* goat udder model. In this study, a simple trajectory was used and no attempt was made to optimize thermal dose delivery. In spite of this simple trajectory, the post-treatment MRI images appeared to show contiguous destruction of the glandular tissue in the ablated volume. There was no evidence of viable tissue within the ablated region. Further, there was no evidence of skin burns or other adverse events that resulted from the MRgFUS ablation procedure. While there are several noteworthy differences between the goat udder and human breast, this translational study has

FIG. 9. Thermal dose accumulated volumes and nonenhancing volumes obtained 20 min after the ablation procedure for animals 2 (a) and (d), 4 (b) and (e), and 6 (c) and (f). Two separate lesions were created in animal 4 (b) and (d). The transducer was located at the bottom of the images in all cases. Volumes that achieved greater than 240 CEM_{43°C} are (a) 1.23 cm³, (b) 2.04 cm³, and (c) 3.60 cm³. Lesion sizes were 8 × 7 × 22 mm (a), 11 × 7 × 14 and 11 × 9 × 13 mm (b), and 14 × 11 × 32 mm (c).

demonstrated the potential of this new system to safely and effectively ablate mammary tissue in a range of udder sizes.

While a goat udder has a shape that is similar to a human breast, there is limited fat present in the udder in both nonlactating and lactating animals, as shown in Fig. 4. While human breast anatomy varies significantly between patients,

there is generally a large quantity of fat distributed throughout the breast, particularly in older patients. Therefore, while a PRF-MR thermometry technique works very well in a goat udder, it would only provide temperature information in the glandular and tumor tissue of a human breast. Implementation of MR thermal mapping techniques using T1 or T2 have

FIG. 10. Scatterplots comparing the nonenhancing volume measured from DCE-MRI and the volume of tissue accumulating 240 CEM_{43°C} (a) Twenty min and (b) 14-days postablation. The linear regression correlation coefficients were measured to be 0.13 and 0.66, respectively. Bland-Altman plots for the ratio of the nonenhancing to thermal dose volumes as a function of the mean of both values (c) 20 min and (d) 14-days postablation. Top and bottom lines indicate the mean ratio ± 1.96 standard deviations with the middle line indicating the mean ratio.

FIG. 11. Histological sample of the side of an excised ablated tissue removed from animal number 2. The transducer would be located at the bottom left corner of the image. A transition region (large arrow) between the normal breast ducts and lobules and adjacent granulation tissue is observed.

shown promise in measuring temperature in fat and could be applied clinically.^{34–36}

Another notable difference between the human breast and the goat udder model is the respiratory effect on temperature measurements. With the human patient in a prone position, the lungs are located directly above the treated breast. Other groups have found that respiration can cause significant temperature errors, as high as 13 °C, when using a PRF based temperature measurement technique.³⁷ In the goat udder, the lungs are located cephalad to the udder. While the respiration effects are evident in the PRF temperature measurements when using a reference-based reconstruction technique, they are much less significant than what would be seen in humans. Therefore, in humans it is likely that an atlas-based method would need to be used either alone or in conjunction with the referenceless reconstruction technique to accurately measure temperature in aqueous tissues using the PRF method.³⁸

The breast MRgFUS system used in this study has a treatable volume of 1100 cm³. Only one mammary gland per animal was targeted with the ablated volume placed either mid-gland or towards the midline of the udder. In addition, there are some space limitations due to the design of the system. The lateral transducer design does increase the height of the system. This may potentially limit the size of patient that may be treated using this device; particularly in a 60 cm bore size. While this limitation will decrease in a larger bore size, it will not be completely eliminated.

While the system had been shown to have excellent repeatability for peak temperature rise in *ex vivo* tissues, there was significant variability in peak temperature rise for equivalent sonications in all animals, as seen in Figs. 5 and 6. Generally, within a sonication group, the first sonication achieved a lower temperature rise than subsequent sonications, as would be expected due to heat conduction. However, the first sonication group generally achieved a higher temperature rise than subsequent groups. It has been shown in other *in vivo* studies that both peak temperature rise and ablated volumes can vary significantly between individual sonications.³⁹ In this case, the variability may have been a function of the

energy delivery method. Power values ranged from 24 to 70 acoustic W applied for 30–60 s at each individual point. Other studies have shown that higher powered, shorter sonication times increase the repeatability of lesion volume by eliminating perfusion dependence.⁴⁰ In addition, most of the goats were lactating and it is known that during lactation the mammary gland exhibits high metabolic and synthetic activity requiring significant changes in the microvascularization.⁴¹ This may have affected both inter- and intraanimal variability in the results. There was also a significant amount of edema present during the ablation period. This tissue swelling could have affected the temperature response of the tissue due to changes in both local tissue properties and ultrasound beam propagation.

In general, immediately after the treatment, the discrepancy between the accumulated thermal dose and nonperfused volume increased with the size of ablation. This phenomenon has been observed by other investigators.¹⁷ This discrepancy may be due to the edema present during the ablation period. In general, the tissue swelling increased with larger ablations. This may have potentially caused changes in local tissue properties and ultrasound beam propagation. There may have also been some bias in the reconstruction method if heating propagated into the reference region, underestimating the temperature measurements. Fourteen days post-treatment, while the agreement between the thermal dose measurement and NPV improved for larger ablations, it decreased for the smaller ablations. This improved agreement may be due to shrinkage of the necrotic volume due to the elimination of dead cells by the body. In addition, the goats were actively nursing kids and the overall udder size changed between the two time points. This tissue volume change may have affected the size of the nonenhancing volume 14-days post-treatment.

In previous studies using this breast MRgHIFU system, it was found that the ultrasound beam placement was both accurate and repeatable.²⁹ In general, lesions were longer in the ultrasound propagation direction. The focal spot of the ultrasound transducer used in this study has a full width half maximum size of 2 × 10 mm, so elongation in the propagation direction was to be expected due to thermal build up in the near field.^{42,43} However, in some cases, the elongation of the ablated volume was larger than expected (animals 6 and 7). Although the tissue was allowed to cool (approximately 4 min) between each sonication group, tissue was ablated in the near- and far-field in some cases. There was interanimal variability in lesion sizes for similar trajectory patterns, as seen in Fig. 9. It is anticipated that the implementation of realtime feedback control and treatment optimization in conjunction with the realtime monitoring of the near- and far-field regions would improve ablated volume control and reduce interanimal variability. Those improvements will be evaluated in a future study.

Nearfield tissue monitoring can be accomplished in 2D imaging by either using slices parallel to the ultrasound propagation path or by placing some slices in both the focal region and the nearfield.^{43,44} In this study, 3D MR thermometry imaging was used with the slab orientated parallel to the

ultrasound propagation path (the thin slab direction perpendicular to the beam path), allowing the operator to monitor the entire acoustic propagation path and immediately identify undesired tissue necrosis or skin heating. 3D imaging also allows data interpolation in all three directions to minimize the under-prediction of temperature rise due to a partial volume effect.⁴⁵ However, there are some disadvantages to 3D imaging, including increased sensitivity to motion and increased acquisition time (4.9 s in this study). The multichannel coil of this system has been shown to allow for parallel imaging with a possible reduction factor of 2×2 (Ref. 30), reducing this potential limitation. While every MRgFUS treatment area has individual temporal and spatial resolution requirements, the temporal and spatial requirements of the 3D MR thermometry sequence used in this study would be feasible to use in a clinical breast MRgFUS treatment.

V. CONCLUSIONS

The *in vivo* evaluation of a new breast-specific MRgFUS system has been performed in a goat udder model. The system was able to successfully ablate lesions up to 3.6 cm³ with no signs of damage to sensitive tissue at the skin surface or chest wall. Tissue damage was confirmed through the correlation of thermal dose measurements obtained with realtime 3D MR thermometry to DCE-MR images immediately after the treatment and 14-days postablation. The thermal dose volume was found to correlate better with the 14-day postablation non-enhancing DCE-MRI image volumes. The necrosis was confirmed qualitatively through histological analysis in one animal. While the goat udder is not an ideal model for the human breast, this study has proven the feasibility of using this system on a wide variety of udder shapes and sizes, demonstrating the flexibility that would be required in order to treat human subjects.

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