A novel MR-guided interventional device for 3D circumferential access to breast tissue

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MRI is rapidly growing as a tool for image-guided procedures in the breast such as needle localizations, biopsy, and cryotherapy. The ability of MRI to resolve small (<1 cm) lesions allows earlier detection and diagnosis than with ultrasound. Most MR-guidance methods perform a two-dimensional compression of the breast that distorts tissue anatomy and limits medial access. This work presents a system for localizing breast lesions with 360° access to breast tissue. A novel system has been developed to perform breast lesion localization using MR guidance that uses a 3D radial coordinate system with four degrees of freedom. The device is combined with a novel breast RF coil for improved signal to noise and rotates 360° around the breast to allow medial, lateral, superior, and inferior access minimizing insertion depth to the target. Coil performance was evaluated using a human volunteer by comparing signal to noise from both the developed breast RF coil and a commercial seven-channel breast coil. The system was tested with a breast-shaped gel phantom containing randomly distributed MR-visible targets. MR-compatible localization needles were used to demonstrate the accuracy and feasibility of the concept for breast biopsy. Localization results were classified based on the relationship between the final needle tip position and the lesion. A 3D bladder concept was also tested using animal tissue to evaluate the device's ability to immobilize deformable breast tissue during a needle insertion. The RF breast coil provided signal to noise values comparable to a seven-channel breast coil. The needle tip was in contact with the targeted lesion in 89% (25/28) of all the trials and 100% (6/6) of the trials with targeted lesions >6 mm. Target lesions were 3–4 mm in diameter for 47% (13/28), 5–6 mm in diameter for 32% (9/28), and over 6 mm in diameter for 21% (6/28) of the trials, respectively. The 3D bladder concept was shown to immobilize a deformable animal tissue phantom during needle insertion. It is concluded that the MR-guidance system accurately localizes small targets on the order of 3–4 mm in a breast phantom with 360° rotational access. © 2008 American Association of Physicists in Medicine. [DOI: [10.1118/1.2952442](https://doi.org/10.1118/1.2952442)]

I. INTRODUCTION

Magnetic resonance imaging (MRI) often provides visualization of breast lesions that escape detection when using traditional mammography or high frequency breast ultrasound (US).^{1,2} Screening with MRI is an increasing trend due to recent clinical studies in high risk patients. Recent evidence on breast MRI screening caused the American Cancer Society to now recommend screening with MRI for women with a significant increased risk of breast cancer as an adjunct to

mammography.³ An observational study by Kuhl *et al.* reported that MRI diagnosed 92% of the surgically confirmed cases of ductal carcinoma *in situ* compared to 56% diagnosed by traditional mammography.⁴ Thus, MRI can be an important tool in the early detection of breast lesions, providing additional opportunities for early diagnoses and various minimally invasive interventional procedures or therapies, such as core biopsy and cryoablation.

Large-core needle biopsy (LCNB) is a minimally invasive procedure to obtain targeted tissue samples. In addition to

being well-accepted by patients and physicians, LCNB has been shown to take less time than surgical biopsy, is cost effective, and is safe.⁵ By allowing the characterization of benign and malignant lesions by histological examination, LCNB is considered the gold standard procedure for diagnosis of palpable and nonpalpable breast lesions.^{6–9} As LCNB gains popularity, the placement accuracy of the biopsy needle will remain a critical issue in retrieving tissue samples reliably and quickly. To guide the placement of the needle into the breast during LCNB, US is commonly used because of its availability, mobility, real-time interactive guidance, and nonionizing radiation. However, US may not be able to identify some MR detected lesions which remain US occult. US also falls short of MRI in needle placement accuracy because it may not have high enough soft tissue contrast to distinguish small lesion boundaries. It is the superior small lesion sensitivity and near real-time capability that has motivated work on MR-guided LCNB and other MR image guided interventional (IGI) breast procedures. Yet, recent results have demonstrated that MR-guided LCNB procedures may not be as accurate as stereotactic or sonographically guided biopsy.¹⁰

To prove the feasibility of MR-guided breast lesion localizations, some groups have reported successful results of needle placement by employing a freehand technique.^{11,12} Other groups developed IGI devices using a fenestrated grid with compression plates for a two-dimensional (2D) approach to localizations.^{13–19} However, the compression plates force the breast into a 2D conformation similar to mammographic plates, which distorts surrounding anatomy, is uncomfortable for the patient, and can lead to the recognized “accordion effect” during needle localization.¹⁴ Although two groups have reported medial access under MR guidance with grid systems that employ 2D compression,^{20,21} these methods often make it difficult, and in some cases impossible, to access lesions from the medial side of the breast because of the geometries of the device and breast coil. There may also be lesions posterior to the access grid during a 2D approach near the chest wall that are inaccessible with the conventional approach.

The purpose of this study is to demonstrate the intrinsic accuracy of MR-guided breast lesion localizations with a novel device for MR-guided interventions in a largely rigid body gel phantom using a true three-dimensional (3D) coordinate system capable of 360° motion around the breast including medial, anterior, and posterior access. In addition, the feasibility of the device for imaging the breast and immobilizing breast tissue was tested in volunteer and phantom experiments.

II. MATERIALS AND METHODS

II.A. IGI device and planning algorithm

A 3D radial approach to breast lesion localization was developed based on a mechanical localization apparatus combined with a 3D reference and control system. The mechanical localization component of the MR-IGI device (hereafter, “the device”) consists of a stationary base that supports

FIG. 1. Device degrees of freedom tested in this study: (θ) Rotation of the platform, needle guide, and coil, (y') elevation of needle guide, (Φ) vertical needle inclination, and (r) depth of insertion. Rotation allows 360° access to superior, inferior, medial, and lateral lesions from a trajectory that minimizes tissue penetration. These settings are assigned by the planning algorithm and are given in the planning output. The calculated trajectory (dotted black line) is the calculated path from the algorithm that minimizes tissue penetration. There is no preferential accuracy for any subset of angles because of the 360° motion. The fifth DOF was not implemented in the algorithm at the time of this study.

a rotating platform (Fig. 1). This platform contains fiducial markers and a needle guide apparatus. The height of the needle guide apparatus above the rotating plane is adjustable and the needle guide can perform inclination about a set pivot point as necessary. These localization components were designed around the natural cylindrical form of a distended breast of a patient lying comfortably prone, which approximates a 3D volume centered on the axis of rotation of the device. This volume is delineated by two primary subvolumes: one, a cylinder where the needle guide rotates around the breast and translates vertically without inclination and two, cones that define the region accessible by an inclined needle guide. Using a 3D planning image acquired to include the target and fiducial markers of the device, the reference and control system is capable of localizing targets within this volume using a fiducial marker system that is referenced to the rotational center of the device. The coil cup and parts of the probe guide were constructed from a rapid prototype process using ABS (acrylonitrile butadiene styrene) while all other parts were either delrin or nylon. All of the materials used in the construction of the device are nonferromagnetic and tested safely in a 1.5 T environment.

Specifically, the device tested in this study improves upon an earlier prototype of the 3D IGI positioning concept that was evaluated using a commercial breast coil.²² Improvements in the new design (Fig. 1) include four degrees of freedom (DOF) for improved flexibility in accessing targeted tissue, a novel solenoid RF breast coil used to both transmit and receive that improves signal to noise and allows medial access,²³ and an improved fiducial marker system in the base to allow more accurate registration of the device geometry to the imaged field of view (FOV) for localization.

During the procedure, the biopsy needle or other interventional probe is inserted from the outer circumference of the 3D volume using rotating and translating polar coordinate settings (Fig. 1). The device allows the needle guide apparatus to rotate circumferentially around the breast. After recording the positions of the fiducial markers and the target, four DOF are calculated by the planning algorithm. The flexibility of the device motions allows the algorithm to minimize the needle path distance through breast tissue, improv-

FIG. 2. Volume rendering in the graphical user interface (GUI) for lesion targeting and trajectory planning. The 3D volume can be rotated freely and orthographic maximum intensity projections (MIP) are displayed (right column) to facilitate trajectory planning. Reference cursors are placed on the device's fiducial markers (white arrows) and the targeted lesion to register the device geometry to the imaging field of view (FOV). Once registration is complete, the trajectory positioning output generated by the planning algorithm is displayed for allowing access to the target.

ing the accuracy of needle positioning. There is a small window of possible trajectories that if not adjusted will encounter the middle loop of the coil. If this occurs, a small adjustment is made automatically to change the elevation of the probe guide and the calculated trajectory is angled vertically to avoid contact with the middle loop. In these cases, the altered trajectory is some small deviation from the minimal needle path distance.

A planning volume is acquired with MRI and transferred to a 3D graphical user interface (GUI) application running on a Linux-based workstation (NVIDIA Quadro 400 NVS graphics card) located next to the MRI scanner host computer. The GUI reconstructs the planning image volume into a 3D rendering that can be manipulated for visualization of the target and intervention trajectory planning (Fig. 2). 2D orthographic projections can also be viewed simultaneously.

From this interface, the positions of the target and four MR-visible fiducial markers are identified within the 3D rendering using visible reference cursors to register the device geometry to the FOV. The planning algorithm uses the positions of these cursors in 3D to set up a coordinate system referenced to the device. The position of the target is then referenced to this coordinate system allowing the algorithm to determine the movement commands using simple geometry for the four DOF tested to define the needle trajectory at the angle that minimizes the through-tissue distance to the target lesion [i.e., horizontal rotation (θ), vertical elevation (y'), vertical inclination (ϕ), and depth of needle insertion (r)].

II.B. Methods to validate device performance

II.B.1. Coil comparison

The capability of the solenoid breast coil was tested using a commercial seven-channel coil (GE 1.5 T Excite 7 Chan-

FIG. 3. Integrated device and MR breast coil mounted within the prototype patient platform used in the localization tests. The breast phantom was placed within the coil from above. The needle guide assembly, as shown by the white arrow, was always positioned lateral at the start of the experiments before the planning volume was acquired. The entire assembly sits on the scanner couch.

nel Breast Array, InVivo Corp., Orlando, FL). Signal to noise (SNR) measurements were compared in a volunteer using a 3D T1-weighted spoiled gradient echo (SPGR) sequence (11.0/4.2/30° TR/TE/FA; 3.0 mm slice thickness; 16.0 cm FOV; 256 × 256 matrix) at different areas within the image. These values were tabulated for comparison. Additionally, the susceptibility of the device was tested with a spherical phantom along all three orthogonal planes using a conventional 2D gradient echo sequence (68.0/4.2/15° TR/TE/FA; 7.0 mm slice thickness; 20.0 cm FOV; 256 × 256 matrix). The phase and frequency encoding directions were swapped and the resulting images subtracted to observe any susceptibility errors.

II.B.2. Localization testing

All tests were performed on a 1.5 T MRI scanner (GE Signa HDx; Milwaukee, WI) using a rigid breast phantom with ten spherical targets of varying size (3–12 mm) and position embedded in a gel matrix (Gammex RMI; Middleton, WI). The targets contained a T1-shortening solution that provided an easy distinction between target and surrounding gel. In order to demonstrate the intrinsic accuracy of both the device and the algorithm, 28 localization attempts were performed over three separate days. Each day began by placing the device within a prototype patient platform on the scanner bed (Fig. 3), placing the breast phantom in the coil, and acquiring one planning image volume using a 3D fast SPGR sequence (11.2/4.2/15° TR/TE/FA; 1.2 mm slice thickness; 20.0 cm FOV; 256 × 256 matrix). Subsequent planning volume reconstruction and localization planning of each target was performed within the GUI. Multiple lesions were targeted per planning volume scan provided the phantom did not move between or during the localizations.

In less than 5 min, the trajectory planning settings from the GUI output were properly set on the device and an MR-compatible breast lesion marking wire (E-Z-EM Corp., Westbury, NY) was inserted manually. A 3D balanced steady-state free precession (SSFP) acquisition (3.1/1.2/30° TR/TE/FA; 1.1 mm slice thickness; 18.0 cm FOV; 160 × 160 matrix) was used for rapid verification of the needle tip position with respect to the target. The 3D image was reformatted to result in a 2D image in the oblique plane containing the long axis of the needle and the lesion to better visualize the trial out-

come. Anisotropic voxel dimensions ($1.125 \times 1.125 \times 1.1$ mm) caused the resolution of the reformatted image to be slightly dependent on the reformat angle.

The results of the verification scans were used to visually assess whether or not the needle tip was in contact with the lesion. Each localization trial was evaluated based on a “contact” or “no-contact” result. If an attempt produced a no-contact result, it was determined if the needle tip offset was parallel to the trajectory path (depth inaccuracy) or perpendicular to the trajectory path (angular inaccuracy). A depth inaccuracy was recorded by using the 2D reformatted image of the long axis of the needle and measuring the distance between the needle tip to the nearest contact point of the target lesion. An angular inaccuracy was recorded by using a 2D reformatted imaging plane through the lesion perpendicular to the needle axis and measuring the perpendicular distance from the target lesion to the needle tip. The size of the lesion was also recorded along with the device setting of the planning output for each trial.

The distribution of contact and no-contact results were evaluated based on target size, depth of insertion, and by location within the phantom to capture the performance for different anatomical quadrants of the breast typically accessed for biopsy (upper medial and lateral, lower medial and lateral).

II.B.3. Tissue immobilization

A 3D immobilization prototype was tested and consisted of three air bladders connected in parallel to a common pressure source (at this prototype stage, a small hand pump). The three air bladders resided within the RF breast coil and allowed an inserted needle to pass through an insertion window. The gel breast phantom used to test the accuracy of the device in localizing lesions was too rigid to represent the expected mobility of a distended breast. Therefore, three chicken breasts were laid in a thin plastic layer to mimic breast tissue mobility. The same fiducial markers used in the device were used as reference markers implanted into the chicken breasts to assess the extent of tissue displacement upon insertion of a polypropylene rod (diameter 3.175 mm) with a trocar tip. The rod was used to mimic MR-guided biopsy procedures performed with a commercially available MR-compatible vacuum assisted biopsy system (Suros, Indianapolis, IN). With the bladders deflated, localization of a target using the planning algorithm was attempted with the rod and subsequent tissue displacement was observed by incrementally inserting 10 mm after tissue contact until the target depth of 77.3 mm was reached. The localization attempt did not require vertical inclination, so after each increment, a 2D coronal image of the trocar was acquired with a conventional 2D gradient echo sequence (68.0/4.2/15° TR/TE/FA; 7.0 mm slice thickness; 20.0 cm FOV; 256×256 matrix). Next, the device was rotated 10° so the next localization trajectory was independent of the first. The bladders were then inflated to less than 1 psi and the localization attempt was repeated for the new trajectory.

FIG. 4. Volunteer images obtained from the solenoid coil (top panel) and the seven-channel coil (bottom panel). Images were obtained with a T1 weighted sequence without fat suppression (left) and with fat suppression (right).

III. RESULTS

The solenoid coil produced diagnostic images of comparable SNR to the commercial seven-channel coil. Examples of the images from the comparison with a human volunteer are shown for the solenoid coil and seven-channel coil (Fig. 4). The SNR values of the solenoid coil and the seven-channel coil are listed in Table I. Moreover, no additional susceptibility errors were observed due to the presence of the device or coil cup when compared to a commercial breast coil.

MR-guided needle localization was performed using the IGI device in 28 attempts at lesions within a breast phantom over three separate days. Of the 28 lesions targeted, 25 (89%) were determined to be a contact result of the needle tip with the lesion. One of the three no-contact results was recorded as an angular inaccuracy with a distance of 2 mm from the lesion surface to the needle tip, corresponding to an angular error of approximately 1°. The second and third no-contact results were determined to be depth inaccuracies measuring distances of 2 and 6 mm, both results short of the target. The maximum insertion depth possible from the geometry of the device (the radius of the cylindrical 3D volume) was 85 mm without needle inclination and was

TABLE I. Signal to noise comparisons for the solenoid coil used in the localization experiments and a clinical seven-channel array coil.

	Coil SNR	
	Solenoid	Seven-channel array
Fat	56.1	71.3
Fat ^a	5.6	7.7
Ductal tissue ^a	20.5	29.8

^aMeasurements taken from the fat-saturated images.

Targeted Lesion Distribution and Results

FIG. 5. Relative locations of the middle of each target with respect to the central axis of rotation are displayed projected onto the axial plane (left) and coronal plane (right). A closed circle represents a “contact” result of the needle tip with the target and an open circle represents a “no-contact” result. The breast phantom consisted of three coronal layers of varying size targets as seen in the axial projection. Target size is not distinguished.

98.1 mm with the maximum needle inclination. One of the two trials that required a needle insertion of 80 mm or more was the no-contact result that was 6 mm short. This outcome is addressed in Sec. IV.

The distribution of contact and no-contact results of the trials is shown as closed and open dots, respectively, in Fig. 5. The three no-contact results were located in the upper medial region of the phantom. Therefore, all three no-contact results required medial access with 0.1° , 0.4° , and 12.0° vertical inclination settings of the needle guide. Examples of trial outcomes are shown in Fig. 6 for no vertical inclination (a), 14.9° vertical inclination (b), and a 2 mm angular inaccuracy no contact result (c).

The contact rate as a function of target diameter and target depth is given in Table II. For the phantom, 47% (13/28) of

the targets were 3–4 mm, 32% (9/28) of the targets were 5–6 mm, and 21% (6/28) of the targets were >6 mm in diameter. The contact rate was actually lower for the intermediate sized, 5–6 mm, diameter lesions as compared to smaller and larger lesions. Lesion depth is reported as the distance from the needle guide pivot point to the target.

Setup for each day of trials took approximately 5 min. The scan time to acquire the planning volume was 6 min and loading in the image volume into the GUI took less than 1 min. To manually place and confirm the position of each reference cursor representing the four reference markers on the device averaged less than 5 min. Planning for each target took less than 1 min and setting the device specified by the GUI took less than 5 min in all cases. Therefore, if only one

FIG. 6. Verification images (3D balanced SSFP) of needle tip placement with respect to the targets (white arrows) demonstrating results without vertical inclination (a) and with vertical inclination (b). A “no-contact” result between needle tip and target is also demonstrated in (c) with a 2 mm angular inaccuracy. A faint white signal band surrounding the phantom in the images is a signal from the breast RF coil’s plastic structure. Initially not intended to produce signal, the faint coil signal provides beneficial visual landmarks during the 3D trajectory planning. The introducer sheath that the localization wire is inserted through is also visible in (a) and (c) as a wider signal void intersecting the phantom periphery.

TABLE II. Contact rate of needle tip with target segregated by target diameter and target depth.

	Contacts/ attempts ($n=28$)	Rate
Target diameter (mm)		
3–4	12/13	0.92
5–6	7/9	0.78
>6	6/6	1.00
Target depth (mm) ^a		
40–50	6/6	1.00
51–60	8/9	0.89
61–70	3/3	1.00
71–80	7/8	0.88
80+	1/2	0.50

^aTarget depth is given as distance from pivot point of the needle guide.

FIG. 7. Relative tissue movement during trocar insertion is shown in the animal tissue phantom without 3D bladders (a) and with 3D bladders (b). Coronal difference images (right) obtained by subtracting the baseline image (left) from subsequent images acquired after incrementally increasing the insertion depth along the planned trajectory (dotted line). Bright or dark signal in difference images demonstrates tissue displacement during insertion relative to the appropriate baseline image. Bright and dark points (white arrows) represent where the nearest markers were located before and after insertion, respectively. The trocar needle was inserted through the insertion window (star) in between compartments b_1 and b_3 .

lesion was targeted for localization, the average time for each procedure would be approximately 23 min using the described method. For each additional target in each session, there would be an average added time of about 6 min because initial setup and device registration have already been performed.

The tissue immobilization tests successfully demonstrated 3D tissue immobilization during a trocar insertion using inflated circumferential air bladders [Fig. 7(b)] compared to an insertion without inflation [Fig. 7(a)]. The pressure used to immobilize the tissue in this experiment was less than 1 psi.

IV. DISCUSSION

The results demonstrate the feasibility and reliability of the planning algorithm and IGI device developed for accurate needle localizations of small lesions in a breast phantom. The 3D method of targeting shows the success of localizing lesions as small as 3 mm within an acceptable time frame.

The three no-contact results in the present study required both medial access and needle inclination. Setting the vertical inclination and controlling the depth of insertion was more difficult during medial access compared to lateral access due to the prototype patient platform. Due to space constraints of the prototype platform, it is likely that the upper medial lesions in this study were subject to a higher potential of human error. Based on these results, the patient platform is currently being redesigned to provide improved visibility for medial access. In addition, optical encoder technology will be integrated into the next iteration of the device to provide easy to read digital output of the positions of all DOF to increase precision of the needle tip positioning and decrease the time required to set the planned trajectory. These improvements would eliminate the human error involved in setting the device, which is believed to be the main reason for the three no-contact results of the study. There may also have been a small amount of error due to trajectory displacement by the gel material in the phantom. Because the wire was very thin, the gel may have displaced the original trajectory and caused a small deviation between the predicted needle tip position from the planning (target) and the

real needle tip position. However, in only one attempt out of 28 was this observed. Biopsies and therapies, such as LCNB and cryotherapy, use much larger diameter and more stable needles or probes, minimizing this concern.

The trial that resulted in the needle tip being located 6 mm short of the target surface was most likely due to manual error in the insertion depth. It should be noted that an insertion that exceeds or falls short of the target depth along the trajectory path can be corrected for by measuring the difference between the needle tip and the target surface in the verification image and adjusting the needle depth accordingly. If an insertion results in an angular inaccuracy, a rotational reposition would have to be made after complete withdrawal of the needle and reinsertion would be needed. However, the results presented for this study are for the first attempt at each target and do not reflect any corrections.

Since for each session only one volume scan was acquired for multiple targets, the accuracy of subsequent localization attempts may have been influenced by previous interventions. For instance, it is likely that the accuracy was degraded by motion between the time that the planning volume was acquired and subsequent interventions because no mechanism existed to secure the device to the table or to connect it to the patient platform. Therefore, the registration of the phantom and the device produced by the GUI before the first intervention of the day may have been altered by subsequent intervention. However, the accuracy achieved despite this supports the ability of the system to biopsy multiple targets with good accuracy using a single planning volume. In the next model of the patient platform, a clamping mechanism will allow the device to be attached securely to the scanner couch. Also, the rigid gel breast phantom used is not an adequate phantom for modeling the distortion of breast tissue expected for *in vivo* needle insertion. However, tissue distortion is outside the scope of the present study, which is designed to determine intrinsic accuracy of the 3D geometry for interventional breast procedures.

There are a number of improvements to the device and procedure that will improve its capability and overall accuracy in target localizations. In order to provide improved access to targets located near the chest wall, both the vertical range of motion and the range of needle inclination will be increased. A fifth DOF, horizontal angulation, is available on the device but was not implemented into the algorithm at the time of the experiments. This fifth DOF is designed to allow small adjustments in needle trajectory to avoid structures or regions that may cause increased trauma for the patient or impede the optimal needle path in the clinical setting.

In order to maximize patient comfort and to preserve the natural anatomy of the breast presented in the diagnostic MR images, an optimized inflatable bladder with a needle insertion window will be incorporated within the coil cup that will be filled with air or a warm saline solution at a suitable pressure needed to immobilize the tissue. This will provide a more distributed pressure than compression plates and will stabilize a variety of breast sizes for interventional procedures. The clinical value of the present device will depend on its capability to immobilize tissue. Preliminary tests have

demonstrated the feasibility of a 3D tissue immobilization technique. These tests validate the accuracy measurements of the device presented in this study. Clearly the next step is to test the accuracy of the device in humans with the tissue fixation bladder.

Although the device was operated with a single solenoid coil for testing, it is intended to be used with another solenoid coil for bilateral diagnostic imaging. The resulting images can be loaded into the GUI and used for planning. The coil with the device can be interchanged with the coil without the device. In order to preserve the immobilization throughout the procedure, the following series of steps outline the steps of immobilization with respect to the MR-guided biopsy procedure:

- Diagnostic acquisition (bilateral)
- Diagnosis and initial target planning (device positioning)
- Immobilization
- Planning acquisition
- Planning refinement (fifth DOF as necessary)
- Needle insertion
- Positional confirmation
- Biopsy, place clip

Although the device setting and biopsy are performed with the patient out of the magnet, imaging is performed with a plastic rod to confirm the needle position. Therefore, the procedural outcome directly depends on the MR guidance. Patient studies are planned to optimize the bladder shape and evaluate the required stabilization pressure. A continuous air bladder may improve performance when compared to the three compartment design and minimize tissue distortion further.

In addition to the design improvements under development, future work will focus on improving the device's performance in localizing breast lesions to prepare for trials in human biopsy procedures. The first objective involves the relocation of the fiducial markers. The markers are currently positioned on the rotating base, which causes the FOV to be larger than the tissue itself requires. The scan time of the planning image volume will be reduced by positioning the fiducial markers closer to the breast tissue along the upper rim of the breast coil, allowing an approximately 50% reduction in the acquired FOV. The markers do not affect the image quality of the tissue as they are not coupled closely to the tissue, either in the current setup or the proposed relocation of the markers. A longer-term objective is to add remote-controlled powered actuators to the DOF and real-time guidance to enable 4D MR-IGI capabilities. In conjunction with robotics, tissue immobilization is an important area of future work so that the best possible accuracy is achieved. Bovine or cadaveric tissue models will be explored for their physical characteristics dealing with needle insertion. The tissue models will be tested in combination with the circumferential bladder previously described to achieve a reliable 3D immobilization technique that is comfortable for the patient.

In summary, a MR-IGI method using a 3D radial localization device and control system, an integrated solenoid breast RF coil, and graphic planning system is described and tested in experiments with a gel phantom containing randomly distributed target lesions. The accuracy of the device was 89% (25/28 lesions contacted) and the precision was better than 6 mm for the lesions tested. These results demonstrate feasibility of this design for MR-guided biopsy and therapy procedures.

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