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Focal Point Determination in Magnetic Resonance-guided Focused Ultrasound Using Tracking Coils

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Abstract

Purpose—Develop a method to rapidly predict the geometric focus location in MR coordinates of a focused ultrasound transducer with arbitrary position and orientation without sonicating.

Methods—Three small tracker coil circuits were designed, constructed, rigidly attached to the transducer housing of a breast MRgFUS system with five degrees of freedom, and connected to receiver channel inputs of the MRI scanner. A 1D sequence applied in three orthogonal directions determines the position of each tracker, which is then corrected for gradient nonlinearity. In a calibration step, low-level heating locates the ultrasound focus in one transducer position orientation where the tracker positions are also known. Subsequent ultrasound focus locations are determined from the isometric transformation of the trackers. The accuracy of this method was verified by comparing the tracking coil predictions to thermal center of mass calculated using MR thermometry data acquired at 16 different transducer positions for MRgFUS sonications in a homogeneous gelatin phantom.

Results—The tracker coil predicted focus was an average distance of 2.1 ± 1.1 mm from the thermal center of mass. The 1D locator sequence and prediction calculations take less than one second to perform.

Conclusion—This technique accurately predicts the geometric focus for a transducer with arbitrary position and orientation without sonicating.

Keywords

MRgFUS; ultrasound focus location; tracker coils

INTRODUCTION

Magnetic resonance-guided focused ultrasound (MRgFUS) has a wide range of promising applications including the treatment of cancer (1-3), localized drug delivery (4-6), and neuromodulation (7,8). For many of these applications the treatment time can be long and may require sonicating at several positions with the transducer in multiple physical location. Thus rapidly finding the physical location of the ultrasound (US) focus is critical to

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successful interventional treatments. When the orientation of the FUS system is monitored with position sensors, the transducer location can be directly computed. Relative to the transducer, the focal spot of table top vertically-shooting transducers have been located by calibrating the transducer focus with gelatin phantoms or finding the water spout in a water bath relative to the MRI scanner isocenter and then calculating offset positions into the patient in MRI coordinates (9,10).

For FUS systems without position sensors, two methods of locating the transducer focus are typically used. In one method, the focus location in MRI coordinates is predicted geometrically from scout images showing the transducer location and orientation. Hand drawn measurements are made to find the focus location based on the known focal length from the transducer face. This location is highly subjective, depending on 1) scout image resolution, 2) the accuracy of the scout image orientation with respect to the transducer, and 3) the accuracy of a perpendicular line drawn from the center of the transducer face.

In a second method, low power short duration heating or acoustic radiation force impulse (ARFI) imaging (11-13) can also be used to determine the focus location. de Bever et al (14) demonstrated the ability of a 3D MR-ARFI sequence to determine the position of the focus to within approximately 0.5 mm in a gelatin phantom using a vertically propagating transducer. To be accurate and sensitive, this method required motion encoding and slice encoding gradients to be along the beam propagation path.

Several factors can contribute to the difficulty in focal spot localization when using low power heating or ARFI. The accuracy of the transducer focus location depends on the acquired image's resolution (15). It has been shown that tissue heterogeneity causes US aberrations which can cause decreased maximum pressure, deformed focal shape, and shifted focal location (16,17). These aberrations increase the difficulty of locating the geometric focus through US sonication. Focus localization with both ARFI and low power heating requires having a reasonable estimate of the focus location from the scout images to effectively position the image volume during sonication. Difficulty of all described methods is increased when the transducer has arbitrary position and orientation.

Dumoulin et al (18) described a simple method for locating a small receive coil in 3 dimensional MR coordinates. This technique has been used extensively in intravascular catheter tracking (19). In a variant of this technique, Ooi et al (20) used three small signal generating beads with individual RF coils placed on a pair of glasses to follow the motion of the head. During image acquisition, tracking data from the beads were used to prospectively update the position and orientation of the imaging volume to follow the motion of the head. The coils on the glasses were wirelessly coupled to the multi-channel head coil, which limited the viable range of motion to correctly distinguish the small receiver coils. Tracker coils have also been incorporated in FUS systems (21-23). In a system designed for prostate ablation (21,23), the tracker coils were placed in the system housing such that the plane formed by the three coils contained the ultrasound focus. A special pulse sequence was used to track the locations of the coils within the imaging volume and to control the image plane to contain the tracker coils in real-time to ensure the focal spot is always within the image.

In this work, we have developed a technique to locate the geometric focus of a focused ultrasound transducer without position sensors, from the locations of three small RF tracking coils mounted rigidly to the transducer housing. Hardware and software design considerations and predicted focal position accuracy relative to US heating measurements are presented.

METHODS

Hardware

All experiments were performed in a Siemens TIM Trio 3T MRI scanner using a breast-specific MRgFUS system (Figure 1) with an integrated eight-channel RF coil and an MRI-compatible phased array transducer (256 elements, 940 kHz frequency, 10 cm radius of curvature; 10 cm focal length; Imasonic, Besançon, France and Image Guided Therapy, Pessac, France). This breast-specific MRgFUS system is the second generation of the system described in (24-26) and is designed to position the patient prone with either the left or right breast in the treatment cylinder at the MRI scanner isocenter. The transducer is mounted laterally to the breast and the system gives the transducer five mechanical degrees of freedom shown in Figure 1b. The entire system can rotate freely about its central axis ($\Theta = 0^\circ$ to 270°) with the patient table in place. Allowed transducer motions include: linear ($\delta = 0$ to 4.5 cm) along the slot guide toward and away from the central axis of the cylinder; tilt ($\Phi = 0^\circ$ to 35°) from horizontal in the plane perpendicular to the MR table; and rotation ($\alpha = -15^\circ$ to 15°) in the horizontal plane of the transducer. These degrees of freedom have gauge markings on the device to display the transducer orientation and aid in positioning the focus at the desired MR coordinates. When combined with electronic steering, the system provides a large treatable volume (0.9 L) within the breast (25).

The three tracker coils were made by tightly wrapping insulated 27 AWG wire four times around a benzonatate 100 mg 6.75 mm diameter spherical capsule. The wires were bonded to the capsule with epoxy and soldered to a custom printed circuit board measuring $21 \times 29 \times 1$ mm (Figure 1d). The unit was rigidly attached to the circuit board with epoxy with the coil axis approximately 15° from vertical to maintain coil signal sensitivity for any Φ rotation of the transducer. The circuit board was printed with a large ground plane on both sides with connection vias to reduce the stray inductance of the circuit. The tuning capacitance for the 1H resonant frequency was achieved by placing capacitors in parallel and in series (figure 1d). Surface mount capacitors were used instead of variable capacitors to ensure capacitance stability over time. A diode and chip RF choke were inserted into the circuit to provide active detuning of the resonant circuit during RF transmission. The coils were connected to Siemens preamps with varying lengths of RG316 coax cable. Preamp decoupling values for the three tracker coils were -22.1 , -24.2 and -23.4 dB. The circuit boards were rigidly attached to the posterior transducer support structure of the breast MRgFUS system forming a triangle with unique side lengths (Figure 1c).

Software

A simple one dimensional readout sequence was used to obtain each tracker coil's approximate position within the bore (Figure 1e) (18). The sequence non-selectively excited

the entire volume then read in one dimension (pixel spacing = 0.24 mm; field of view = 500 mm; echo time (TE)/repetition time (TR) = 4.6/9.4 ms; readout bandwidth = 250 Hz/pixel; flip angle = 15°). After applying a Hann filter, the one-dimensional data was zero-fill interpolated to 0.12 mm pixel spacing. The sequence used six excitations to perform a readout in all three orthogonal directions with both a positive and negative gradient lobe. The data was converted to image space by the scanner image calculation environment (ICE) and transferred automatically (through the ultrasound control software) to a custom MATLAB (Mathworks Ind, Natick, MA) graphical user interface (GUI) on a second computer, which performed all of the following computations. A center of mass (COM) was computed for each readout polarity using Eq 1 where r_i is the position of each voxel and v_i is the associated signal value:

$$COM = \frac{\sum_{i=1}^n v_i (\vec{r}_i) \cdot \vec{r}_i}{\sum_{i=1}^n v_i (\vec{r}_i)} \quad [1]$$

Only voxels whose signal magnitude exceeded 15% of the maximum were included in the COM calculations. The final tracker coil position was computed from the mean of the two center of mass locations for each dimension (18).

The coil positions were then corrected for gradient nonlinearity, also referred to as gradient warp, using the method described by Janke et al (27). The spherical harmonic coefficients necessary for the correction were provided by Siemens Medical Solutions, Erlangen, Germany. After obtaining the position of each coil, the six degrees of freedom transformation (28) that aligned the calibrated coil locations to the current locations was calculated using MATLAB's *procrustes()* function. This transform was then applied to the calibrated focus location to estimate the current focal location. The MATLAB GUI displayed the current focal location, current coil positions and the raw signal plots from each coil. The computation time within the GUI was approximately 150 ms. This prediction method requires calibration of the tracker coil positions with respect to the focus location. The calibration is described below in the Accuracy Verification section.

Tracker Coil Signal

An experiment was performed to assess the performance of the tracker coil signal as a function of distance from isocenter. The worst case scenario for signal performance was chosen where the MRgFUS system was placed in the scanner with the transducer's beam propagating in the head-foot (HF) direction, as this alignment of the tracker coils with the B0 field has the highest sensitivity to Φ rotations (Fig 1b). The center tracker coil (coil #2, Fig 1c) on the transducer housing was placed at isocenter, and the MR patient table was moved in 1 cm increments out to 20 cm away from isocenter while collecting tracker location data at each table position. This was repeated for four Φ rotations (0°, 10°, 20°, and 30°). The magnitude of the gradient warp correction distance of the tracker coils was also calculated for these data. The focus was predicted at each HF position, both with and without using gradient warp correction, using the same calibrated coil positions and focus

obtained from a separate experiment as described below in the Accuracy Verification section.

Accuracy Verification

A breast shaped gelatin phantom (29) was placed in the system and coupled to the transducer with deionized and degassed water that was doped with 1 g/L of manganese chloride (MnCl_2) to decrease T2 and suppress the water signal. The center of the MRgFUS system tank was positioned at isocenter and the system patient table was placed above the phantom and six one liter bags of saline were placed on the table above the transducer to emulate signal from a patient. The transducer was placed in a total of 16 positions, of which 12 were unique and 4 were repeated. With the center of the MRgFUS system positioned at isocenter, the tracker coils were located approximately 16-20 cm from isocenter. This distance will vary slightly depending on which breast is treated and system rotation Θ . The gelatin phantom was heated with the US transducer at the geometric focus (13 acoustic Watts, 31.5 seconds) twice for each transducer position where the 3D imaging slabs were oriented with the imaging planes parallel and perpendicular to the US propagation path (Figure 2). Proton resonance frequency (PRF) shift temperature measurements (30) were obtained during sonication using a three dimensional segmented echo planar imaging (seg-EPI) sequence with flyback readout (voxel dimensions = $1 \times 1 \times 2$ mm; field of view = $224 \times 154 \times 16$ mm; TE/TR = 19/41 ms; EPI Factor = 13; 12 slices with 25% oversampling; readout bandwidth = 1062 Hz/pixel; 13 Image repetitions; 6.3 s per image). After applying a Hann filter along the readout and phase encoding directions, images were zero-fill interpolated to 0.5 mm isotropic voxel spacing. Three baseline images were acquired before US sonication, and the average phase during the baseline images was used as the reference phase for calculating the PRF shift temperature change.

The measured focus location in both slab orientations was determined as the COM of the 3D temperature map using only the voxels that experienced at least 50% of the maximum temperature. The measured focus location for each transducer position was defined as the average of the locations determined from the parallel and perpendicular imaging slabs. The temperature COM at the first transducer position (Figure 2a with $\Theta=0^\circ$, $\delta=2\text{cm}$, $\Phi=10^\circ$, $\alpha=0^\circ$) was used to calibrate the location of the focus with respect to the three tracker coils. As the calibration between the tracker coils and the focus was not yet known, the 3D imaging slabs were aligned manually for the first location by estimating the focal position from measurements drawn on localizer images to ensure the focus was captured in the images. After the calibration from the first location was determined, predictions from the tracker coils were used to position the center of the slab at the predicted focus location for the remaining transducer positions. The slab orientation (rotation) was aligned parallel or perpendicular to the US propagation path with knowledge of the rotation, Θ , of the system and the rotations (Φ , and α) of the transducer. The distance between the measured and tracker coil predicted focus locations was determined for each transducer position.

The variability of the measured temperature COM position was measured by repeating the heating at the first transducer position six times (3 parallel slabs and 3 perpendicular slabs) while allowing sufficient time for the gelatin to cool between repetitions. The tracking

sequence was repeated 10 times at the first transducer position to assess the variability of tracker coil position estimates. The focal position was predicted from each of these 10 repetitions.

RESULTS

Figure 3a shows the typical signal from a tracker coil for all 6 readouts. The dashed line shows the cutoff set at 15% of the maximum value. Signal from the saline bags, which is well below the 15% cutoff, can be seen in the Y direction. The measured coil position is defined to be halfway between the peaks from positive and negative readouts. The typical relative signal-to-noise ratio (SNR) of these coils with the tracking sequence was approximately 1000. Figure 3b shows the tracker signal as a function of head-foot distance from isocenter and Φ rotation. The signal change with Φ rotation depends on the component of the coil's area that is perpendicular to the MRI's main magnetic field. The kink in the plot is possibly due to the profile of the B1 transmit field and the related bandwidth of the excitation pulse, though this has not been explored. Figure 3c shows the gradient warp correction distance as a function of head-foot distance from isocenter for a tracker coil on the magnet's $x=0$ (left-right) axis. Figures 3d-e show how the predicted focus moved with and without gradient warp correction compared to the first predicted location for the left-right (LR) and anterior-posterior (AP) axes for $\Phi=30^\circ$. The predicted focus should show no motion along these axes, as the only change between these points was the table moving in the HF direction. Figure 3f shows the deviation from expected difference in the predicted focus position relative to the first predicted location along the HF axis for $\Phi=30^\circ$.

Temperature measurements as a function of time for the six repeatability sonications were nearly identical within the parallel (3.1% variation) and perpendicular (2.0% variation) cases. The parallel slab orientation measured a slightly higher temperature than the perpendicular slab, most likely due to partial volume effects (15). For the parallel slab repeatability tests, the temperature center of mass moved an average distance of 0.15 mm between repetitions, and for the perpendicular slabs it was 0.14 mm. The average distance of the temperature center of mass between the parallel and perpendicular imaging slabs was 2.1 ± 0.1 mm in the repeatability tests, and for all 16 locations it was 1.6 ± 0.5 mm.

The tracker coil predicted focus was an average distance of 2.1 ± 1.1 mm from the locations determined from the PRF temperature data. Without gradient warp correction the predicted focus was an average distance of 6.5 ± 3.9 mm from the PRF temperature focus. The error in predicted focus location displayed some bias. The average error in the y axis prediction was approximately 1.4 mm above (Patient Posterior) the actual focus. The average error in the x-z plane was approximately 1.1 mm towards the transducer. For the 10 run tracker coil repeatability tests, the measured center of mass positions moved an average distance of 9 micrometers, and the position of the predicted focus varied by an average distance of 19 micrometers between runs.

Figure 4 shows the PRF temperature change at the ultrasound focus and predicted focus location overlaid on the zoomed in magnitude images for all three orthogonal planes going

through the temperature center of mass for all 16 transducer positions of the parallel imaging case.

DISCUSSION

The results in this paper demonstrate that, using tracking coils without sonication, the geometric focus of the US transducer in this specific non-commercial MRgFUS system inside a Siemens TIM Trio 3T MRI scanner can be accurately predicted to within approximately 2.1 mm. The calibration between the tracker coils and focus locations only needs to be performed once for the system. The tracking sequence and prediction calculations are rapid, requiring less than one second to measure and report the predicted focus in MR coordinates. Such rapid localization might also be applicable to a hand held or manually positioned FUS transducer. While this rapid position determination is very useful to FUS therapies, the need for low power test sonications may still be necessary to ensure proper acoustic coupling of the patient as well as other safety and efficacy concerns. However, the tracker coils will minimize the time necessary for focal point determination. Gradient warp correction proved to be necessary to accurately locate the coils. This is due to their relatively large distance from isocenter (~16-20 cm) for most transducer positions.

Indeed, imperfect gradient warp correction is most likely the largest source of error between the predicted and actual focus location. Gradient warp correction uses a limited number of correction coefficients, and requires accurate measurements of the spherical harmonic coefficients. Figures 3e-f show the effects of imperfect gradient warp correction on the predicted focus. While significantly improved after gradient warp correction, more correction coefficients could possibly further improve the accuracy of the tracker coil predicted focus. Tracker coils implemented on ultrasound systems with shorter focal lengths may have less gradient warp error, as they would likely be closer to isocenter during treatment. The tracker coils' effective range is limited to the size of the magnet's spherical volume of uniform field around isocenter and the usable range of the gradients, which will vary between scanner models.

The procrustes algorithm used in MATLAB calculates a shape preserving Euclidean transformation. By making the triangle shape formed by the three tracker coils rigid and unique, i.e. not isosceles or equilateral, it ensured that the correct transducer translation and, more specifically, rotation is calculated. The exact placement of the trackers on the transducer housing is not necessarily important as long as the triangle sides are unique and they move rigidly with the transducer. While MATLAB offers a convenient prototyping environment, future development plans include performing all calculations within the ultrasound control software, removing the need to export to MATLAB.

Tuning and matching of the coils provided high SNR, allowing the coils to be used anywhere within the magnet's uniform volume. The high SNR also makes the position determination highly repeatable. It should be noted however, that proper decoupling of the tracker coils during image acquisition is necessary to prevent artifacts in normal imaging sequences. When active/passive or preamp decoupling fails, the tracker coils can produce artifacts in imaging sequences even when the coil is not within the excited volume or

imaging field of view. Quality assurance scans with phantoms, already in the treatment protocol before each patient is treated, will ensure the tracker/imaging coils are in proper working order.

The repeatability of heating with FUS in the gelatin phantom was demonstrated. The predicted focus location does not take into account any US propagation effects such as refraction, beam aberration or attenuation. For each transducer position, the US beam propagated through different distances in the phantom. These different propagation lengths would attenuate the focus slightly, helping explain the discrepancy between the heated and predicted focal location since the predicted location is dependent on geometry only. This shift is likely much smaller than the error from imperfect gradient warp correction due to the homogenous nature of the phantom. Finally, it is possible that beam aberration distortions may be greater in breasts with highly heterogeneous distributions of fat and glandular tissue (29).

CONCLUSIONS

This work demonstrates that the positions of three tracker coils rigidly attached to the transducer housing can be used to quickly and accurately predict the location of the ultrasound transducer geometric focus without the need to sonicate. Rapid prediction of the focus based on this method will shorten total treatment time by allowing faster focal spot determination. This will improve patient safety and potentially reduced treatment times by removing the need to sonicate in order to locate the focus.

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FIG 1.

a) MRgFUS breast system b) Cross section showing various degrees of freedom. c) Tracker coil arrangement. The transducer is cut in half horizontally to clearly show tracker coils. d) Conceptual circuit design. Coil axis is rotated $\sim 15^\circ$ from vertical to give maximum signal possible for every transducer position. e) Locator Sequence for one readout direction G_R . This is applied along G_x , G_y and G_z to locate the tracker coil.

FIG 2.

Transducer positions and imaging slabs. Green – Parallel slab. Yellow – Perpendicular slab. RO – Readout direction. a) Initial position at $\Phi=10^\circ$, $\delta=2$ cm, $\alpha=0^\circ$. b) $\Phi=20^\circ$, $\delta=2$ cm, $\alpha=0^\circ$. c) $\Phi=20^\circ$, $\delta=3$ cm, $\alpha=0^\circ$. d) $\Phi=20^\circ$, $\delta=3$ cm, $\alpha=10^\circ$. Red lines show approximate ultrasound beam propagation. Curved blue line outlines the transducer face. All experiment locations used one of these four positions with the system rotation Θ at 0° , 40° or 90° . See figure 1b for definitions of Θ , δ , α and Φ .

FIG 3.

a) Example tracker signal from a single coil for all 6 readout directions with inset showing zoomed in profiles. b) Tracker signal vs head-foot distance from isocenter for four Φ rotations for tracker coil #2. c) Gradient warp correction distance as a function of head-foot distance from isocenter for four Φ rotations for tracker coil #2. d-e) Predicted focus movement relative to the first predicted location with and without gradient warp correction for the LR axis (d) and AP axis (e). f) Deviation from expected difference relative to the first predicted focus location with and without gradient warp correction for the HF axis.

FIG 4.

PRF temperature at the focus overlaid on magnitude images for the three orthogonal planes through the temperature center of mass for all 16 transducer positions of the parallel slab imaging case. The calculated temperature center of mass location is shown by a green marker and the predicted focus location from the tracker coils is shown by a blue marker.