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Real-time probe tracking using EM-optical sensor for MRI-guided cryoablation

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Abstract

Background—A method of real time, accurate probe tracking at the entrance of the MRI bore is developed, which fused with pre-procedural MR images will facilitate clinicians to perform the cryoablation efficiently in large workspace with image guidance.

Methods—Electromagnetic (EM) tracking coupled with optical tracking is used to track the probe. EM tracking is achieved with an MRI-safe EM sensor working under the scanner's magnetic field to compensate the line-of-sight issue of optical tracking. Unscented Kalman filter-based probe tracking is developed to smooth the EM sensor measurements when the occlusion occurred and to improve the tracking accuracy by fusing two sensors measurements.

Results—Experiment with a spine phantom show the mean targeting errors using the EM sensor alone and the proposed method are 2.21 mm and 1.80 mm, respectively.

Conclusion—The proposed method achieves more accurate probe tracking than using the EM sensor alone at the MRI scanner entrance.

Keywords

MRI-guided cryoablation; electromagnetic tracking; optical tracking; probe tracking; unscented Kalman filter

I. Introduction

Cryoablation in which probes are placed into the tissue to destroy cells by freezing [1], has been used in the treatment of spine tumors and pain palliation [2]–[4], as it provides minimal trauma and results in a low complication rate [5]. The success of cryoablation is highly dependent upon intraoperative imaging for tracking the probe in the anatomy.

Several imaging modalities have been used in this procedure: ultrasound [6]–[8], fluoroscopy [9], computed tomography (CT) [2], [4], [10]–[12], and magnetic resonance imaging (MRI) [13], [14]. MRI safely provides the greatest soft tissue contrast [15]. However, the confined physical space in a close-bore MRI scanner (diameter, approximately 50–70 cm, length 125–200 cm) presents a significant challenge for clinicians to access to the patient in the bore. The patient must be repeatedly moved into the bore for imaging so as to verify targeting accuracy and out of the bore for probe adjustment, which can be time consuming and fatiguing to both the patient and the clinician. This repeated operation increases the procedural time, which may take more than two hours. Even though the open MRI scanner has more operation space, which allows interactive guidance with near real-time imaging feedback [16], [17], its field strength is lower and its image quality is inferior to that of the closed-bore MRI. In addition, it is not widely available.

Accurate identification of the probe's tip in MR images is another complicated task for the clinicians, who estimate the probe tip's position in the anatomy based on the probe's artifact in MR images. The large artifact makes it difficult for clinicians to localize the probe's tip accurately. Inaccurate probe placement may result in damage to the nearby anatomy and sub-optimal ablation margin. To improve the accuracy of probe tracking, Song et al. studied the method of probe localization according to its artifacts in MR images [18], [19]. Shimizu, et al. used pattern recognition algorithms to identify the probe [20], which is usually not robust due to the anatomical occlusion in MR images. Diamagnetic coatings to enhance the probe's artifacts in MR images have been used for more accurate detection of the probe's tip [21], [22]. However, these methods require frequent MR scanning and rather heavy image processing. A method in which a sensor is affixed to the probe to track its position can reduce the scanning times and avoid the heavy image processing, but few sensors are available due to the MRI incompatibility.

In this study, a method for real-time accurate probe tracking with an MRI-safe electromagnetic (EM) sensor and optical sensor at the entrance of the MRI's bore is proposed and validated. Optical tracking is the most popular method for tracking the surgical instruments in the operation room as it provides accurate measurements, although there is a line-of-sight issue. The MRI-safe EM sensor originally designed for working under the gradient magnetic fields at the isocenter of the scanner is used to address this issue. However, the EM sensor can be prone to noise and inaccuracy while tracking the probe at the MRI entrance due to the highly nonlinear gradient fields. Unscented Kalman filter was developed to improve the probe tracking accuracy using both EM and optical sensors. A phantom experiment was performed to validate the proposed method.

II. Materials and Methods

A. Probe with EM-optical sensor

The EM sensor plus the optical sensor (called “EM-optical sensor”) are attached to the probe. Fig.1 shows the configuration of the probe (length 175 mm; diameter 1.5 mm; IceRod® 1.5 MRI, Galil Medical, Arden Hills, Minnesota, USA) for cryoablation with the EM-optical sensor. The probe is made of nickel-chromium superalloy and composed of a cone tip, a shaft, and a handle. There are tick marks on the probe's shaft to facilitate the

clinicians determining the depth that the probe is inserted into the tissue. Both the EM sensor and a nylon frame equipped with four retro-reflective markers for optical tracking are affixed to the probe's shaft. The offset from the probe's tip to the sensor can be adjusted according to the tick marks.

The EM sensor is an MRI-safe, six degrees-of-freedom EM sensor (EndoScout sensor, Robin Medical Inc., Baltimore, MD, US) [23–25] with the tracking accuracy of 2 mm and 1 degree in terms of position and orientation. In order to maintain MR compatibility, the EndoScout sensor can only work in the gradient magnetic fields generated by the MRI scanner using a special sequence. It is composed of six sensing coils. If a coil with a cross-section area A and the unit normal $\mathbf{n}$ is placed at the position $\mathbf{P}$ in the gradient magnetic field, which is assumed to be approximately uniform and represented by $\mathbf{G}(t, \mathbf{P})$, the voltage V induced in the coil can be calculated according to the Faraday's law as

$$V = - \frac{d(\mathbf{G}(t, \mathbf{P}) \cdot \mathbf{n}A)}{dt} \quad (1)$$

Six coils are arranged on the six cubic planes in the EndoScout sensor to track the sensor's position and orientation in the gradient magnetic field. The voltage induced in the coils can be measured and the gradient magnetic fields are known. Therefore, the position and orientation of the sensor can be determined by solving six differential equations corresponding to each coil. As the same physical phenomenon (gradient magnetic fields) is used to reconstruct the MR image, the tracking is performed in the same coordinate system as the imaging. Therefore, there is no need for coordinate system alignment and registration.

The nylon frame equipped with four retro-reflective markers are tracked by an MR-safe optical tracking device (XINOMDT, Symbow Medical Technology Co., Ltd., Beijing, China) designed by Symbow Medical based on Polaris Spectra (Northern Digital Inc., Canada). Its specifications are the same as those of Polaris Spectra with 0.25 mm RMS in the range from 950 mm to 2400 mm.

Multiple coordinate systems are used to map the probe's tip motion (see Fig.2): the optical coordinate system is denoted by Σ_{Opt} , the EM coordinate system by Σ_{EM} , and the MR image coordinate system by Σ_{Img} .

In the probe intervention procedure, the clinician maneuvers the probe according to the location of the probe's tip in the anatomy so as to navigate the probe to the target. However, the position of the probe's tip cannot be measured directly as an offset from the sensor/markers to the probe's tip exists. The offset from the origin of the rigid body defined by the four markers to the probe's tip can be estimated using pivot calibration [26]. After the calibration, the probe's tip position, denoted by $\mathbf{p}_{opt}^{tip}$, can be obtained from the optical device directly. For EM tracking, two clamps on one side of the sensor are designed to hold the probe shaft. The measurements of the sensor consist of the position of the intersection point O (denoted by $\mathbf{p}_{EM}^O$) between the probe shaft's axis and the right plane of the right clamp

(see Fig. 1), and the orientation of the probe shaft's axis (denoted by $\mathbf{r}_{EM}$). Let x denote the offset from $\mathbf{p}_{EM}^{tip}$ to $\mathbf{p}_{EM}^O$. The position of the probe's tip in Σ_{EM} is then computed as

$$\mathbf{p}_{EM}^{tip} = \mathbf{p}_{EM}^O + \Delta x \mathbf{r}_{EM} \quad (2)$$

The offset x can be adjusted by moving the sensor along the probe shaft according to the tick marks on the probe shaft. In this study, x is set at 100 mm.

B. Spatial transformation and calibration

The probe's tip is tracked in different coordinate reference systems: Σ_{Img} , Σ_{EM} , and Σ_{Opt} (see Fig.2). To visualize the probe's trajectory in the pre-procedure MR images, the measurements acquired by the two sensors should be converted in to a common coordinate system. In this work, the MR coordinate system is chosen as the common coordinate system. Therefore, two spatial transformations should be estimated in advance, i.e., $^{Img}\mathbf{T}_{Opt}$ (from Σ_{Opt} to Σ_{Img}) and $^{Img}\mathbf{T}_{EM}$ (from Σ_{EM} to Σ_{Img}). As there is no deformation involved, only rigid transformation is considered.

The spatial transformation $^{Img}\mathbf{T}_{Opt}$ can be estimated using fiducial marker-based registration, which is reliable and widely used in the field of image guided surgery [27], [28]. Seven fiducial markers on an MRI phantom is used, three on the left side plane and four on the cylinder surface, see Fig.3(a). To localize these markers in the MR images, the phantom was moved into the isocenter of the gantry for scanning. T1-weighted MR images (image spacing: 1.023 mm×1.023 mm×1.5 mm) are obtained and imported into 3D Slicer [29]. The center of each marker's basin is manually localized by the user according to three-orthogonal slice views (see Fig.3(b)), and its coordinate in Σ_{Img} , denoted by $\mathbf{p}_{Img}$, is recorded. The phantom is then moved out of the isocenter by 650 mm to the entrance. Each marker's position in Σ_{Opt} is acquired by the optical tracking device when the probe's tip attaches to the center of marker's basin, defined as $\mathbf{p}_{Opt}$. After all markers' positions in Σ_{Img} and Σ_{Opt} are collected, $^{Img}\mathbf{T}_{Opt}$ can be estimated by solving the following equations using the least squares method.

$$\hat{^{Img}\mathbf{T}_{Opt}} = \underset{^{Img}\mathbf{T}_{Opt}}{\operatorname{argmin}} \| \mathbf{p}_{Img} - ^{Img}\mathbf{T}_{Opt} \cdot \mathbf{p}_{Opt} \| \quad (3)$$

The MR images and the EM sensor refer to the same magnetic field, Σ_{EM} is assumed to be identical to Σ_{Img} , meaning that $^{Img}\mathbf{T}_{EM}$ is an identical matrix. Nevertheless, the movement of the phantom (or patient) from the isocenter to the entrance should be taken into account. A

translation should be applied to the measurement of the EM sensor. The homogeneous

representation of the transformation matrix ${}^{Img}\mathbf{T}_{EM}$ is
$$\begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 650 \\ 0 & 0 & 0 & 0 \end{bmatrix}.$$

The EM sensor is originally designed for application inside the bore of the MR scanner. At the entrance of the scanner bore, it may be subject to inhomogeneous gradient magnetic fields, resulting in lower accuracy than that at the isocenter. The EM sensor should be calibrated in the space at the entrance. A bias error map of the EM sensor is created using the optical measurements as the gold standard. Supposing N positions are recorded by the EM sensor and optical sensor respectively and converted into the image coordinate system Σ_{Img} , denoted by $\{(\mathbf{p}_{opt}^i, \mathbf{p}_{EM}^i) | i = 1, 2, \dots, N\}$, let $\mathbf{b}$ denote a bias vector field of the EM sensor. $\mathbf{b}(\mathbf{p}_{EM}^i)$ represents the displacement vector from $\mathbf{p}_{EM}^i$ to $\mathbf{p}_{opt}^i$, i.e., $\mathbf{b}(\mathbf{p}_{EM}^i) = \mathbf{p}_{opt}^i - \mathbf{p}_{EM}^i$. The bias vector field is interpolated using the Radial Basis Function (RBF) interpolation. With the given sample points, the bias vector at an arbitrary point $\mathbf{p}_{EM}$ can be calculated by

$$\mathbf{b}(\mathbf{p}_{EM}) = \sum_{i=1}^N \lambda_i \Phi(\|\mathbf{p}_{EM} - \mathbf{p}_{EM}^i\|) \quad (4)$$

where λ_i represents the coefficient and Φ is the kernel function. The coefficients can be estimated by solving the linear equations with the sampling points as the input. Here, thin plate spline is selected as the kernel function.

C. Unscented Kalman filter-based probe tracking

As the accuracy of the EM sensor is greatly affected by the noise and other factors, such as the velocity, Unscented Kalman filter (UKF)-based probe tracking is developed to improve the accuracy. UKF is an extension of Kalman filter [30], [31], and often used to estimate the state of a nonlinear system due to its high computation efficiency and accuracy [32].

The state vector, denoted by $\mathbf{x} \in \mathbb{R}^6$, describes the position and the velocity of the probe's tip in Σ_{Img} .

$$\mathbf{x} = [\mathbf{p}^T \quad \dot{\mathbf{p}}^T]^T \quad (5)$$

where $\mathbf{p} = [x \ y \ z]^T$ is the position vector, and $\dot{\mathbf{p}} = [\dot{x} \ \dot{y} \ \dot{z}]^T$ is the linear velocity.

The speed of probe insertion is always very slow allowing for the safety, therefore the velocity can be assumed to be constant in the consecutive sampling interval. Suppose that the initial orientation of the probe is $\mathbf{n}$, which points to the target and can be measured by the EM sensor, the motion of the probe's tip can be predicted by

$$\mathbf{x}_k^- = \begin{bmatrix} \mathbf{p}_k^- \\ \dot{\mathbf{p}}_k^- \end{bmatrix} = f(\mathbf{A}, \mathbf{x}_{k-1}, \mathbf{n}, \Delta t, \mathbf{w}_k) = \underbrace{\begin{bmatrix} \mathbf{I} & \Delta t[\mathbf{n}\cdot] \\ 0 & [\mathbf{n}\cdot] \end{bmatrix}}_{\mathbf{A}} \underbrace{\begin{bmatrix} \mathbf{p}_{k-1} \\ \dot{\mathbf{p}}_{k-1} \end{bmatrix}}_{\mathbf{x}_{k-1}} + \underbrace{\begin{bmatrix} \frac{\Delta t^2}{2} \\ \Delta t \end{bmatrix}}_{\mathbf{w}_k} \quad (6)$$

where $\mathbf{A}$ is the processing matrix of UKF, $[\hat{\mathbf{n}}\cdot]$ is an operator to calculate the projection of the velocity vector onto the initial orientation vector $\mathbf{n}$, i.e., $[\mathbf{n}\cdot]\dot{\mathbf{p}} = (\dot{\mathbf{p}} \cdot \mathbf{n})\mathbf{n}$, $\mathbf{w}_k$ is the processing noise, which is associated with the acceleration and assumed to be white Gaussian noise with zero mean and covariance matrix $\mathbf{Q}$.

The measurement vector $\mathbf{z}$ in UKF is the position of the probe's tip, which can be acquired by both the EM and optical sensor (no occlusion) or only the EM sensor (occlusion). The discrete-time measurement model of UKF can be written as

$$\mathbf{z}_k = h(\mathbf{H}, \mathbf{x}_k^-, \mathbf{b}_k, \mathbf{v}_k) \Leftrightarrow \begin{cases} \begin{bmatrix} \mathbf{p}_{EM} \\ \mathbf{p}_{Opt} \end{bmatrix} = \underbrace{\begin{bmatrix} \mathbf{I} & \mathbf{0} \\ \mathbf{I} & \mathbf{0} \end{bmatrix}}_{\mathbf{H}} \underbrace{\begin{bmatrix} \mathbf{p}_k^- \\ \dot{\mathbf{p}}_k^- \end{bmatrix}}_{\mathbf{x}_k^-} + \begin{bmatrix} \mathbf{b}_k \\ \mathbf{0} \end{bmatrix} + \underbrace{\begin{bmatrix} \mathbf{v}_k^{EM} \\ \mathbf{v}_k^{Opt} \end{bmatrix}}_{\mathbf{v}_k} & \text{no occlusion} \\ \mathbf{p}_{EM} = \underbrace{\begin{bmatrix} \mathbf{I} & \mathbf{0} \end{bmatrix}}_{\mathbf{H}} \underbrace{\begin{bmatrix} \mathbf{p}_k^- \\ \dot{\mathbf{p}}_k^- \end{bmatrix}}_{\mathbf{x}_k^-} + \mathbf{b}_k + \underbrace{\mathbf{v}_k^{EM}}_{\mathbf{v}_k} & \text{occlusion} \end{cases} \quad (7)$$

where $\mathbf{H}$ is the measurement matrix, $\mathbf{b}_k$ is the bias error of the EM sensor at $\mathbf{z}_k$, $\mathbf{v}_k^{EM}$ and $\mathbf{v}_k^{Opt}$ represent the measurement noise of the EM and optical sensor, respectively. The measurement noise is also assumed to be white Gaussian noise, i.e., $p(\mathbf{v}_k) \sim \mathcal{N}(\mathbf{0}, \mathbf{R})$, where $\mathbf{R}$ is the covariance matrix of the measurement noise. For the optical sensor, $\mathbf{v}_k^{Opt}$ is provided by the vendor, whereas, $\mathbf{v}_k^{EM}$ is estimated by a linear combination of two components

$$\mathbf{v}_k^{EM} = \begin{bmatrix} \sigma + \omega \log(|v_x| + 1) \\ \sigma + \omega \log(|v_y| + 1) \\ \sigma + \omega \log(|v_z| + 1) \end{bmatrix} \quad (8)$$

where σ is the measurement variance of the EM sensor; v_x , v_y and v_z are the velocity of the probe's tip along each axis measured by the EM sensor; ω , a coefficient balancing the variances of two sensors, is set to 0.5 empirically. Non-linear relation of the covariance with the velocity will increase the dynamic performance of the EM sensor accuracy.

UKF contains two recursive steps: prediction and correction. For each time of data acquisition, these two steps execute once.

1. Prediction:

- ◆ Calculating $2N+1$ sigma-points based on present state covariance:

$$\mathbf{x}_{0,k-1} = \mathbf{x}_{k-1} + \xi(i)\gamma\mathbf{S}_i, \quad \xi(i) = \begin{cases} 0 & i = 0 \\ 1 & 0 < i \leq N \\ -1 & N < i \leq 2N \end{cases}$$

where $\mathbf{S}_i$, $i = 1, \dots, N$ is the i th column of the covariance matrix $\mathbf{P}_{\mathbf{x}_{k-1}}$, i.e., $\mathbf{S} = \sqrt{\mathbf{P}_{\mathbf{x}_{k-1}}}$; γ is a scaling parameter, $\gamma = \sqrt{N + \lambda}$, $\lambda = \alpha^2(N + \kappa) - N$. Here, α and κ are tuning parameters. κ must be a non-negative parameter to guarantee the semi-positive definiteness of the covariance matrix, and is zero in default. The parameter α , $0 \leq \alpha \leq 1$, controls the size of the sigma-point distribution and, ideally, it should be a small number, $\alpha = 0.001$.

- ◆ Transform the sigma points using Eq.(6):

$$\mathbf{x}_{i,k|k-1} = f(\mathbf{A}, \mathbf{x}_{i,k-1}, \mathbf{n}, \Delta t)$$

- ◆ Estimate the prior state:

$$\bar{\mathbf{x}}_{k|k-1} = \sum_{i=0}^{2N} (w_m^i \mathbf{x}_{i,k|k-1})$$

- ◆ Compute the prior error covariance matrix

$$\mathbf{P}_{\mathbf{x}_k}^- = \sum_{i=0}^{2N} w_c^i (\mathbf{x}_{i,k|k-1} - \bar{\mathbf{x}}_{k|k-1})(\mathbf{x}_{i,k|k-1} - \bar{\mathbf{x}}_{k|k-1})^T + \mathbf{Q}$$

- ◆ The weights w_m^i and w_c^i are defined as,

$$w_m^0 = \frac{\lambda}{N + \lambda}$$

$$w_c^0 = \frac{\lambda}{N + \lambda} + (1 - \alpha^2 + \beta)$$

$$w_m^i = w_c^i = \frac{1}{2(N + \lambda)}, i = 1, \dots, 2N$$

where β is a non-negative weighting parameter introduced to affect the weight of the zeroth sigma point for the calculation of the covariance. It can be used to incorporate knowledge of the higher order moments of the distribution. For a Gaussian prior the optimal choice is $\beta = 2.0$.

2. Correction:

- ◆ Transform the sigma points using Eq.(7) according to whether there is occurrence of the occlusion.

$$\hat{\mathbf{z}}_{i,k|k-1} = \mathbf{h}(\mathbf{H}, \mathbf{x}_{i,k|k-1}, \mathbf{b}_k)$$

- ◆ Calculate the mean and covariance of the measurement vector

$$\bar{\mathbf{z}}_{k|k-1} = \sum_{i=0}^{2N} (w_m^i \hat{\mathbf{z}}_{i,k|k-1})$$

$$\mathbf{P}_{\hat{\mathbf{z}}_k^-} = \sum_{i=0}^{2N} w_c^i (\hat{\mathbf{z}}_{i,k|k-1} - \bar{\mathbf{z}}_{k|k-1})(\hat{\mathbf{z}}_{i,k|k-1} - \bar{\mathbf{z}}_{k|k-1})^T + \mathbf{R}$$

$$\mathbf{P}_{\hat{\mathbf{x}}_k^- \hat{\mathbf{z}}_k^-} = \sum_{i=0}^{2N} w_c^i (\mathbf{x}_{i,k|k-1} - \bar{\mathbf{x}}_{k|k-1})(\hat{\mathbf{z}}_{i,k|k-1} - \bar{\mathbf{z}}_{k|k-1})^T$$

- ◆ Compute the Kalman gain

$$\mathbf{K}_k = \mathbf{P}_{\hat{\mathbf{x}}_k^- \hat{\mathbf{z}}_k^-} \mathbf{P}_{\hat{\mathbf{z}}_k^-}^{-1}$$

- ◆ Estimate the posterior state $\hat{\mathbf{x}}_k$. Here, $\mathbf{z}_k$ is the measurement vector, which can be $\mathbf{p}_{EM}$ if there is occlusion, otherwise, $\begin{bmatrix} \mathbf{p}_{EM} \\ \mathbf{p}_{Opt} \end{bmatrix}$ according to Eq.(7).

$$\hat{\mathbf{x}}_k = \bar{\mathbf{x}}_{k|k-1} + \mathbf{K}_k(\mathbf{z}_k - \bar{\mathbf{z}}_{k|k-1})$$

- ◆ Estimate the posterior error covariance matrix

$$\mathbf{P}_{\mathbf{x}_k} = \mathbf{P}_{\hat{\mathbf{x}}_k^-} - \mathbf{K}_k \mathbf{P}_{\hat{\mathbf{z}}_k^-} \mathbf{K}_k^T$$

UKF plays two roles: a smooth filter and a data fusion filter. When there is the issue of line-of-sight for optical tracking, only the EM sensor works. UKF is a smooth filter reducing the

noise of the EM sensor. On the other hand, when there is no occlusion, both sensors work. UKF as a data fusion filter fuses two sensors' data to improve the accuracy.

III. EXPERIMENTS

The proposed method was implemented using C++ as a stand-alone application, and a loadable module for visualization in 3D Slicer. The application runs on a client responsible for data collection and estimation of the probe's tip position; the loadable module in 3D Slicer runs on a server in charge of visualization. The client communicates with the server via Ethernet using an open network protocol, OpenIGTLink [33].

A. Experimental setups

The experiment was performed in the Advanced Multimodality Image Guided Operating (AMIGO) suite¹, a state-of-the-art medical and surgical research suite that houses an array of interventional surgical systems and advanced imaging equipment. Fig.4 shows the experiment's setup, including a 3.0T closed-bore MRI scanner (Verior® 3T, Siemens Medical Solutions USA, Inc.). The MRI-compatible optical tracking device is deployed at the end of the table approximately 2000 mm from the entrance. Due to the crowded scenario, the incidence of occlusion for optical tracking is expected to increase dramatically.

Pivot calibration for optical tracking was performed only once to estimate the offset between the reflective markers and the probe's tip. The max root mean square error was 0.5 mm. When a new probe was obtained, the frame with the markers was attached to the shaft at the same tick mark as that in the pivot calibration. A time-varying, three-dimensional gradient magnetic field for the EM sensor was generated by activating the gradient coils of the MRI scanner using a predetermined activation pattern [23].

A spine phantom was used to validate the proposed method. The phantom is a plastic box filled with a mixture of gelatin and agar, into which a lumbosacral spine model was embedded. Six fiducial markers were attached to the surface of the box.

B. Bias error map of the EM sensor

The EM sensor was originally designed for application within the bore, and its accuracy at the entrance of the scanner's bore was still unknown, and may be affected by the inhomogeneity of the gradient magnetic field. In this work, the EM sensor was calibrated with respect to the optical data as the gold standard. It is reasonable to move the probe randomly in the region-of-interest so as to collect data for calibration. However, it is difficult to do so free-hand. A rubber cube with a probe inserted into it was moved from one square to another on a grid paper overlaid on the scanner's table at the entrance. At each square, the cube was kept static for 10 seconds and both the EM and optical sensors' data were collected. Once the cube reached all squares on the paper, the cube height was increased by approximately 10cm so as to sample on a higher plane in the same manner. Data collection was performed on a total of three planes. The sample range in relation to the MRI coordinate system was from -151.08mm to 161.59mm along x axis, from -71.78mm to 90.85mm along

¹<http://brighamandwomens.org/research/amigo/default.aspx>

y axis, and from -579.15 to -774.12 along z axis. Fig.5(a) illustrates the sample points in an anatomical coordinate system (called RAS referred by 3D Slicer), which is a right-hand system describing the standard anatomical position of a human. Its R(ight)-L(eft) line, A(nterior)-P(osterior) line, and I(nferior)-S(uperior) line are parallel to x-axis, y axis and z axis of the MRI coordinate system, respectively. A bias error map of the EM sensor was generated using Eq.(4) and represented by a grid shown in Fig.5(b).

C. Validation

A retrospective analysis was performed to evaluate the accuracy of the proposed method. First, high resolution T1-weighted MR images of the phantom were obtained (see Fig.6(a)). The phantom was then moved out of the bore by 650 mm and stopped at the entrance of the scanner's bore. Spatial transformation was performed according to the procedure described in Section II-B. Seventeen trials of probe insertion directed toward different targets were performed. In each trial, both the EM and optical measurements were recorded. Following each trial, the probe remained in the phantom and the EM and optical sensors were detached from the probe and affixed to a new probe at the same tick marker as the first one for a new trial. Upon completion of all trials (Fig.6(b)), CT images of the phantom with all probes (see Fig.6(c)) were obtained to locate the position of the probes' tip as the golden standard for CT images higher spatial resolution and less distortion than MR images. In the experiment, one trial was excluded because the probe's position changed prior to CT scanning. The remaining 16 trails were used for the validation process. Both MR and CT images were imported into 3D Slicer. Fiducial markers-based registration was performed to register the CT images to the MR images. The position of the probes' tip were manually labeled on the CT images (see Fig.6(d)) and transformed to the coordinate of the MR images in Σ_{Img} as the ground truth of the target.

IV. Results

The targeting error was calculated as the distance from the measured/estimated position of the probe's tip to the corresponding ground truth. Without the method of UKF-based probe tracking, the targeting error is 4.23 ± 2.51 mm with optical tracking alone and 2.21 ± 1.06 mm with EM tracking alone. To simulate the occlusion, only the EM sensor measurements are provided to UKF in Eq.(7). UKF filters the noise in the measurements and the targeting error is 2.16 ± 1.09 mm. On the other hand, if no occlusion occurs, both EM and optical measurements are fed into Eq.(7). UKF fuses the measurements of the two sensors and the targeting error is 1.87 ± 0.92 mm. Fig. 7 illustrates the targeting error mean and standard deviation.

V. DISCUSSION

MRI-guided cryoablation has advantages over other image modalities' guidance because of its superior soft tissue contrast and arbitrary plane imaging. Currently, this procedure is performed by freehand in a repetitive manner as described in Section I due to the limited size of the bore [34]. Navigation with accurate, real-time probe tracking plus pre-procedural MRI is more efficient in guiding the probe to the target, as there is no need for repetitive image acquisition and verification of the probe's tip. Although the acquisition time for a

high resolution planning volume requires considerable time, the overall procedure time is expected to be substantially lower, because only a single pre-procedural image is required. In addition, the relative large MR volume can accommodate probe intervention for multiple targets.

In this study, an MRI-safe EM sensor was used for probe tracking and its accuracy was poor due to its sensitivity to the MRI gradient magnetic field, which is highly non-linear at the scanner entrance. An error map of the EM sensor at the scanner entrance was generated and used for bias error correction. The error map shown in Fig.5(b) illustrates that the EM sensor error increases from superior to inferior and from medial to lateral due to the gradient field inhomogeneity outside of the bore.

According to the specifications of two types of sensors, the optical tracking accuracy is greater than that of EM tracking. On the contrary, our results illustrated that the accuracy of EM tracking was greater than that of optical tracking, an error of 2.21 mm vs. 4.23 mm, respectively. Because the measurement of the probe's tip position using both EM tracking and optical tracking is based on the assumption that the probe is rigid during intervention. However, in reality, the probe is flexible and may deflect [35]. The probe bending will exaggerate the error between the tracking position and the actual position of the probe's tip and the error is in proportion to the offset from the probe's tip to the sensor. The offset from the optical sensor to the probe's tip is larger than that of the EM sensor, so the optical sensor probe's tip tracking error is greater than that of the EM sensor. The cryoablation probe used in this demonstration is semi-rigid, with a cone tip, whose symmetric shape can avoid the side force acting on the tip. Therefore, its deflection could be controlled much more easily than other biopsy probes. In Case #3–#6, #9–#10, #14–#17 (10 of all cases), the probe was kept straight during the intervention, and the mean targeting error was lower (1.30 mm). In the other cases, the deflection of the probe was greater and the mean error was greater (2.81 mm). It proves that controlling the probe deflection within a narrow range is possible. Nevertheless, compensation for the probe deflection using the optical tracking and EM tracking in combination with a mechanical model will be part of our ongoing work.

There is no significant difference between the targeting error for EM tracking alone with and without UKF-based probe tracking. Nevertheless, UKF could smooth the path tracked by the EM sensor. Fig.8 shows that there is significant bias error along the path of the probe's tip tracked by the EM sensor at the scanner's entrance. The trajectory (blue points) measured by the EM sensor alone has deviated from the real path at a large extent (in the yellow ellipse). The variation of the velocity calculated from the EM sensor measurements indicates that a large bias error is usually accompanied by a high velocity. Based on this observation, the measurement noise was designed as a log function of velocity to reduce this deviation (see Eq.(8)). The estimated trajectory (red points) was smoothed using the proposed method.

Overall, the proposed method of UKF-based probe tracking achieved the aim of compensating for the EM sensor error, which reduced the targeting error to 1.87 ± 0.92 mm by fusing the measurements of two sensors. The targeting error may be affected by several factors, such as image resolution, registration error (including probe calibration error, fiducial maker localization error, spatial transformation error, CT-MR registration error),

error of the sensors, probe deflection, etc. There is no study states an ideal targeting accuracy should be for cryoablation. Zamecnik et al. reported automated real-time needle tracking for MR-guided transrectal prostate biopsy and the mean target accuracy was 1.7 ± 0.8 mm [37]. Eva et al. showed a mean targeting error of 1.8 ± 0.9 mm (in vitro) and 2.9 ± 1.0 mm (in vivo) in their freehand MR-guided percutaneous needle intervention [38]. In addition to the targeting error, the size of the tumor is another factor that the clinicians should take into consideration prior to cryoablation. It has been advised that the probe must be placed approximately 1 cm from the tumor margins and 2 cm from each other [39], [40]. MRI-guided cryoablation is recommended to treat larger lesions: its main advantage of precise visualization of the ablation zone ensures coverage of the total tumor while sparing neighboring healthy structures. The proposed probe tracking method integrated with pre-procedural MRI can be used to guide the probe intervention.

To our best knowledge, this is the first study on probe tracking with an MRI-safe EM sensor plus optical sensors. The method proposed in this study has several advantages. First, the EM sensor shares the same coordinate system with MR imaging, which avoid the registration between EM tracking and MR images. Second, EM tracking can compensate the issue of line in sight of optical tracking, while optical tracking can improve the targeting accuracy with the UKF-based sensor fusion. Third, the proposed probe tracking method combined with pre-procedural MR image is proper to guide the probe intervention, because the patient remains in the same position in the pre-procedural MR scanning and the following probe intervention. The pre-procedural MR images delineate the anatomic status more accurate. MR scanning times can be reduced, which let the intervention more efficient. Last, probe intervention performed at the scanner entrance is an ergonomic solution to that in the limited size bore (see Fig.4).

One limitation of this study is that the phantom material is nearly homogeneous, whereas, human tissue is heterogeneous. One drawback of the probe tracking method is the constant noise from the scanner when generating the gradient magnetic field for EM tracking.

VI. CONCLUSION

A method for real time probe tracking using EM and optical sensors at MR scanner entrance has been developed for overcoming the problem of limited space in the MRI bore. An MRI-safe EM sensor and optical sensors were used to track the probe for fast localization of the probe's tip at the scanner entrance. A UKF-based probe tracking method was developed to improve the accuracy of the EM sensor with the EM sensor's error map for compensation and demonstrated a reduction in terms of the mean targeting error from 2.21 mm to 1.87 mm. The proposed method shows promise in contributing to improve the tracking accuracy and guidance of probes for cryoablation.

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Fig.1.
Probe with EM-optical sensor.

Fig.2.
Multiple coordinate systems.

Fig.3. Localization of fiducial markers: (a) phantom with seven attached markers, two of which were occluded by the phantom body; (b) manual localization of the markers in 3D Slicer.

Fig.4.
Experiment setup in AMIGO MRI room.

Fig.5.

Generation of the EM sensor's error map: (a) sample points by optical tracking (red) and EM tracking (blue); (b) grid representation of the error map. Color bar shows the error in *mm*.

Fig.6. Validation using CT as the ground truth. (a) Axial MR image of the phantom; (b) Phantom with 17 probes; (c) Axial CT image corresponding to (a) after registration; (d) Ground truth of the tips' position (red points) shown in volume rendering mode.

Fig.7.

Targeting error using optical sensor only (Opt), EM sensor only (EM), EM sensor with UKF (EM-UKF), and EM-Optical sensor with UKF (EM-Opt-UKF).

Fig.8. Noise affects the measurement of the EM sensor in Case #16. Blue points represent the trajectory measured by the EM sensor alone and red points are the estimated trajectory using the proposed method.