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Voltage-based Device Tracking in a 1.5 Tesla MRI during Imaging: Initial validation in swine models

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Abstract

Purpose—Voltage-based device-tracking (VDT) systems are commonly used for tracking invasive devices in electrophysiological (EP) cardiac-arrhythmia therapy. During EP procedures, electro-anatomic-mapping (EAM) workstations provide guidance by integrating VDT location and intra-cardiac-ECG information with X-ray, CT, Ultrasound, and MR images. MR assists navigation, mapping and radio-frequency-ablation.

Multi-modality interventions require multiple patient transfers between an MRI and the X-ray/ultrasound EP suite, increasing the likelihood of patient-motion and image mis-registration. An MRI-compatible VDT system may increase efficiency, since there is currently no single method to track devices both inside and outside the MRI scanner.

Methods—An MRI-compatible VDT system was constructed by modifying a commercial system. Hardware was added to reduce MRI gradient-ramp and radio-frequency-unblanking-pulse interference. VDT patches and cables were modified to reduce heating. Five swine cardiac VDT EAM-mapping interventions were performed, navigating inside and thereafter outside the MRI.

Results—Three-catheter VDT interventions were performed at >12 frames-per-second both inside and outside the MRI scanner with <3mm error. Catheters were followed on VDT- and MRI-derived maps. Simultaneous VDT and imaging was possible in repetition-time (TR) >32 msec sequences with <0.5mm errors, and <5% MRI SNR loss. At shorter TRs, only intra-cardiac-ECG was reliable. RF Heating was <1.5C°.

Conclusion—An MRI-compatible VDT system is feasible.

Introduction

Tracking with Electromagnetic fields; magnetic-based and voltage-based tracking

Electromagnetic (EM) tracking of the position and orientation of interventional medical devices was developed in the 1990s to rapidly follow multiple devices inside the body without X-ray radiation. Two main methods emerged: Magnetic field-based tracking¹ and Voltage-based tracking². Both were rapidly adopted into the cardio-vascular (CV) interventional workflow. In addition, EM tracking was used to create 3D anatomical models, by recording both physiologic voltages and the location of the catheter tip each time it touches the walls of blood vessels or heart chambers. Integration of 3D geometric information with intra-cardiac electrocardiograms (ECGs), known as electro-anatomic mapping (EAM)³, is central to modern cardiac diagnostic and therapeutic Electro-physiology (EP) procedures.

Magnetic-based device tracking (MDT)¹, such as the CARTO™ EAM system (Biosense-Webster, Diamond Canyon, CA) utilizes multiple electromagnetic transmitters which are placed underneath the patient table. The transmitters emit spatially decaying kHz-band EM waves, which are received by ferromagnetic-core solenoids placed on the shafts of interventional devices. The amplitudes of these waves are used to locate the position and orientation of each solenoid on the device. The device's shape is then reconstructed by interpolating between points. MDT positional accuracy is typically $1 \times 1 \times 1 \text{ mm}$ ³⁴. Its response is relatively linear within the human body, but its geometric accuracy is affected by the presence of external metallic objects, as well as by dynamic magnetic fields. With MDT-tracked devices, separate solenoids and electrodes are placed on the device's shaft to measure the positional and EGM data, so most devices carry only 1–2 positional probes.

Voltage-based Device Tracking (VDT) systems², such as the EnSite™ Velocity™ cardiac mapping system (EnSite Velocity System, St. Jude Medical, St. Paul, MN, employing EnSite™ NavX™ navigation and visualization technology) uses a set of surface patches placed on the patient's skin (Figure 1). Voltage differentials are set up between pairs of patches using sine waves in the kHz range. VDT utilizes standard EP catheters which typically have multiple electrodes along their shaft, so that the voltage measured between an electrode and a patch provides a measure of the distance between that electrode and that patch. VDT relies on the ionic conductivity of the tissues, and if the ionic conductivity between the patches and the catheter is relatively constant, then VDT positional accuracy and precision is typically $\sim 1.5 \times 1.5 \times 1.5 \text{ mm}$ ³⁵. If the current path goes around low conductivity tissues (e.g. the lungs), however, then significant geometric distortions can be present and accuracy is lower. Consequently, VDT-based EAM models are distorted relative to CT or MRI renditions. Nevertheless, many clinicians favor VDT relative to MDT, since VDT EAM maps are rapidly created, utilizing catheters with multiple electrodes, with each electrode providing both positional and intra-cardiac ECG (EGM) information. An important advantage of VDT is that the ionic currents stay within the body, making VDT less sensitive to external metallic objects and magnetic fields. These VDT advantages are important in the case of positional tracking within the confined space of an MRI scanner, since the MRI cryostat is metallic and there is a strong magnetic field throughout.

Imaging Modalities integrated with EAM systems in the EP suite

Many EP procedures center around the EAM workstation, which can provide a presentation comprised of various imaging data combined with the EAM catheter position and cardiac visualization data. Pre-acquired MRI and CT images are incorporated into the display after co-registration to the EAM frame-of-reference^{6–8}. Intra-cardiac ultrasound (ICE) catheters, used for real-time imaging during the EP intervention, have recently been integrated with EAM^{9–13}. Robotic catheter manipulators have also been integrated¹⁴. In general, integrating new devices with EAM requires the placement of VDT/MDT positional sensors on the device.

Value of MRI guidance in EP procedures for the ablative treatment of arrhythmia

MRI is increasingly used prior to EP intervention for planning¹⁵, as well as afterward, to verify electrical-circuit isolation¹⁶. MR Angiography images are often used to map cardiac chamber lumens. Late Gadolinium Enhancement MR images provide details on myocardial scar arising from both infarction and radio-frequency ablation (RFA)^{15–20}. There is also evidence that post-RFA myocardial edema can be seen with T2-weighted imaging²¹. As a result of MRI's potential to improve EP-procedure outcomes, EAM workstations are increasingly used to import, register, and incorporate MRI data, making it available to the interventionist during the procedure. However, incorporation of MRI data is sometimes problematic^{22–24}, due to co-registration issues caused by patient positioning changes between the MRI table and OR table, or due to undetected motion occurring between imaging and intervention.

Several groups are exploring use of intra-procedural MRI for EP^{7, 15, 25–30}. Soft-tissue visualization during device manipulation can potentially reduce complications during navigation. MR is also able to monitor therapy delivery, improving the accuracy of RF ablation delivery, so that lesion depth is sufficient (trans-mural) and gaps are avoided^{15, 17, 18}. Monitoring may reduce recurrence due to inadequate treatment, or complications due to over treatment, which is significant because recurrence is 30–40% for paroxysmal atrial fibrillation and 50% for ventricular tachycardia^{31–33}, with 5% complications^{34–37}.

Intra-procedural MRI is currently inhibited by the absence of MRI-compatible devices that work both inside and outside the MRI. Although active MRI-based device tracking methods (MR-tracking³⁸ and MRI-gradient-tracking³⁹) exist they only work inside the MRI scanner and require specific device additions. With most EP devices not approved for use in the MR environment, performing the entire clinical procedure inside the MRI is not yet possible, particularly for those procedure stages which must be, or are preferably, performed outside the MRI scanner. Examples of stages which must be performed outside the MRI are guide-wire navigation in meandering and branching blood vessels, trans-septal punctures between the atria, and placement of catheters into the coronary sinus.

Benefits of Voltage-based-device-tracking in X-MR guided EP procedures

In order to exploit the MRI advantages, without being restricted by current limitations, there is an increasing use of EP suites that also include an MRI scanner at close proximity. Such

sites, termed X-MR suites, have a transport mechanism to move the patient from the X-ray-based interventional area to the MRI scanner^{7, 22, 40, 41}. Transport remains cumbersome, since all non-MRI-conditional devices need to be removed from the patient prior to movement. Motion between modalities disrupts procedure workflow, since MRI-data co-registration on the EAM system is repeated each time the patient is returned to the X-ray-based interventional setting.

If VDT could be rendered MRI-compatible, device tracking could be performed reliably both inside and outside the MRI. Devices could be left inside the patient during movement to the MRI, permitting accurate and rapid registration of MRI data with the EAM system. An additional benefit of a universal tracking method would be the ability to easily integrate other modalities into the X-MR suite. ICE catheters equipped with tracking could be rendered MRI-safe and used both outside and inside the MRI scanner, improving navigation, and allowing performance of combined ultrasound and MRI sequences, such as acoustic radiation force imaging or focused-ultrasound ablation.

The project goal was to develop a non-MRI based tracking method that works both inside the MR environment and in conventional EP labs^{42, 43}. VDT was chosen due to its reduced sensitivity to static magnetic fields and nearby metallic structures. Previous attempts to make MDT MRI-compatible have failed⁴⁴, due to problems encountered with MDT distortion caused by metal in the MRI cryostat, as well as with the ferromagnetic-core solenoid sensors that are used in MDT.

Methods

Issues with intra-MRI use of Voltage-based Tracking

Figure 2A is a schematic diagram of a Voltage-based-Device-Tracking system placed in the MRI suite. A VDT system measures two signals from each catheter electrode (channel). The first is the positional localization signal (LOC), which is measured with a kHz-range signal burst. The second signal is the intra-cardiac electrogram (EGM) measured on the cardiac wall by the electrode. The EGM signal ranges between 0–500 Hz. The two signals, which are carried over the same cable, are separated after the VDT amplifier. As mentioned previously, the static magnetic field of the MRI does not strongly couple with the positional signal. It does affect the EGM⁴⁵, however, by creating a magneto-hydrodynamic (MHD) voltage on the EGM, primarily during the ST segment of the cardiac cycle.

In the presence of an MRI imaging sequence, considerable noise is added to both the LOC and EGM⁴⁵ signals. This effect is demonstrated in LOC signals recorded in a human in the presence of a gradient-echo sequence. The LOC signal measured without imaging (Fig. 2B) is composed of serial bursts in the three localization directions, which are corrupted during imaging (Fig. 2C). This interference is due to (I) induced voltages arising from the MRI magnetic-field gradient pulses, as well as (II) from strong (kilovolt), “RF unblanking” pulses used in the system’s body coil during radio-frequency (RF) pulse transmission to reverse bias diodes during spin excitation. The radio-frequency imaging pulses themselves, which are at 64 MHz, do not interfere with the VDT EGM or LOC signals. As seen in Figure 2C, the interference frequently exceeds the peak voltage accepted by the VDT amplifier,

resulting in “clipped” signals. Signal clipping complicates the recovery of VDT signals, and prevents device tracking during scanning. Furthermore, the saturated amplifiers require a considerable amount of time to recover before they start recording true signals once again. To reduce this effect, we implemented a circuit to block signal reception during noisy periods, (Fig. 2D) while recording signals during periods where noise is not present (Fig 2D, left side).

Additional challenges for safe VDT within the MRI room include: A) induced voltages on the VDT patches due to the gradient ramps, which create eddy currents and reduce the MRI SNR in the proximity of the patches, as well as inducing local heating, and B) RF induction into the catheters and the cables that transmit LOC/EGM signals from the electrodes to the room’s penetration panel. Both issues may potentially cause heating at the catheter/tissue and patch/skin surfaces, which if severe, can result in burns.

The MRI-compatible Voltage-Based-Device Tracking System

An EnSite Velocity System was modified for MRI-compatibility with a GE (Waukesha, WI) 1.5 Tesla CV/i MRI running 12× cardiac software on an 8-channel cardiac coil. Within the MRI room, the commercial surface patches were slit at several points along their length (Figure 3A), and the 100cm leads were replaced with 5cm leads ending in 5mm diameter female copper “banana” connectors. These changes were performed to reduce eddy currents in the patches and to reduce the length of single-conductor transmission lines. Single-conductor cables leading from the patches to the VDT amplifier were replaced with thin coaxial cables (Fig. 3B), terminated with a male “banana” connector. Outside the MRI’s magnetic field, ferrite chokes were placed on the cables at 64-MHz half-wavelength distances to attenuate RF propagation. The coaxial shields were connected to the penetration-panel via high-voltage (2KV) millifarad capacitors, preventing 0–60Hz leakage currents, so that both high-pot and leakage current tests were within the regulatory specifications for interventional device use. 20 MHz low-pass filters on the outside of the penetration panel prevented RF noise from entering the room.

Once outside the room, the coaxial cables passed through switching circuitry (Figure 3C). A combined low-pass filter/delay-line was positioned before the electronic switches, delaying the VDT signals by 70 microseconds and reducing the frequency content to <15 kHz. Single-pole-double-throw (SPDT) switches were used to disconnect VDT signals from the amplifier during noisy periods in the pulse sequence.

A Gradient and RF-unblanking circuit detected the noisy periods. Gradient waveforms and RF unblanking signals from the MRI’s system cabinet passed through detection circuits (Figure 3C, lower row) which output a TTL signal during each gradient ramp and RF unblanking pulse. No attempt was made to differentiate between the effect of the three gradient coils, or set a limit on the minimal slew rate to trigger on each, although there is theoretical and experimental evidence^{46–47} that the three coils create electric field disturbances of a varying magnitude. A logical OR circuit combined these TTL signals to create a VDT blanking pulse, which could then be extended in time. The resulting TTL signal was sent to the SPDT switch. The 70-microsecond VDT delay time compensated for

delays in the TTL switching pulses generated by the triggering circuit, preventing transients from getting through the circuit.

In addition to the blocking circuit, the EnSite Velocity System included software under which LOC data, corrupted by noise, was ignored by the input amplifier, and catheter position was maintained at the last known location until uncorrupted data was received. This algorithm was designed for compatibility with MediGuide™ Technology⁴⁸, and required MRI scans with repetition-time (TR) >32ms duration, since this time was required for the VDT input amplifiers to recover from saturation.

St. Jude Medical provided two types of deflectable EP catheters. To compare VDT to MR-Tracking^{28, 30}, we utilized an 2.6 mm diameter non-fluid-irrigated MRI-compatible catheter that had two VDT electrodes at the tip and four MR-Tracking micro-coils along its shaft (Figure 3D). This dual MR- and VDT- tracked catheter was previously described and extensively tested for MRI-compatibility^{28,30}

A second type of 2.2 mm diameter fluid-irrigated MRI-compatible deflectable catheter with four or eight VDT electrodes along its shaft was used for the animal experiments (Figure 3E). It had a non-ferrous stainless steel braid proximal to the catheter's distal 60 mm end, with nitinol braiding used in the distal 60 mm, as well as nitinol pull-wires. The distal 4-mm tip electrode and the more proximal 2-mm width ring electrodes, spaced at increments of 3, 6, 3, 8, 8, 8, and 8 mm, were constructed of Gold."

Testing the MRI-compatible VDT system in Phantoms

The MRI-compatible VDT system was initially tested in a wet-lab phantom (Figure 4A–C). Seven VDT patches, the same number used in humans and animals, were placed on the surface of a cylindrical plastic container (Fig. 4A). A left atrial (LA) model was placed inside the container (Fig. 4B), and the system filled with saline. Dual-VDT and MRI-tracked catheters were inserted into the wet-lab (Fig. 4C) and manipulated to simulate an EP procedure. LA maps were constructed as shown in Figure 4D. *In these experiments, the hardware blocking circuit described previously was not used.*

Two MRI-compatible EP catheters were simultaneously navigated to various points in the phantom. Measurements were made inside the MRI scanner in three cases: A) VDT-based navigation without the MRI pulsing, B) VDT-navigation while the MRI acquired low-bandwidth MRI-tracking spoiled-gradient-echo (GRE) sequences (GE Global Research, Niskayuna, NY³⁸), so that catheter position was also observed on the MR-tracking display, and C) VDT-navigation while the MRI acquired short TR, high-bandwidth 2D or 3D GRE imaging sequences. MR-tracking and VDT accuracy was evaluated at several frames-per-second (fps) rates. The active-visualization platform³⁸ also allowed acquisition of real-time MRI images at the tip of the catheter during the navigational process, utilizing MR-tracking position and orientation data which was sent to the MRI scanner from the tracking interface.

MRI imaging sequences tested; A) Low-Bandwidth: 2D GRE, TR/TE/Θ=17 msec/4 msec/30 degrees, 256×128, 30×30 cm FOV, 5 mm slice width, ±32.5 kHz bandwidth, 2.2 sec/image,

B) High Bandwidth: 2D GRE, TR/TE/Θ = 5 msec/2 msec/30 degree, 256×128, 30×30 cm FOV, 5 mm slice width, ±128 kHz bandwidth, 1 average, 0.6 sec/image.

MR-Tracking sequences³⁸ tested; C) 10 fps high-resolution : ±16 kHz bandwidth, TR/TE/Θ = 5 msec/2 msec/5 degrees, 256 pts, 35 cm FOV, 300 mm slice thickness, 10 excitations/location, consisting of 3 phase ditherings (120 degree increments) in each spatial direction and a reference projection. D) 20 fps low resolution: ± 32.5 kHz bandwidth, TR/TE/Θ = 5 msec/2 msec/5 degrees, 128 points, 35 cm FOV, 300 mm slice thickness, 4 excitations/location, consisting of a single phase dithering in each spatial direction and a reference projection.

Animal Models

Swine models (N=6) were used to test the safety (N=1) and efficacy (N=5) of VDT navigation inside the MRI suite, with navigation performed first inside the MRI bore, and then outside the MRI bore on the MRI table. Six healthy 25–45 Kg swine were used under an animal-committee approved protocol. All animals were intubated in the animal laboratory, and mechanically ventilated. Venous access large enough for insertion of 3mm catheters was obtained in a jugular vein and in the bilateral femoral veins. 3mm femoral artery catheter access was obtained on one side. Transport to the MRI suite was performed on a dolly. In the MRI suite, the animals were placed on the MRI table and continuously monitored with an MRI-compatible physiological monitoring system (Invivo Expression, Gainesville, FL) and ventilated using an MRI compatible veterinary ventilator (DRE Premier XP, Louisville, KY), with forced breath-holds performed by shutting off the ventilator. Other experimental details have been previously described^{28, 30}. Swine heart rates ranged from 70–100 BPM.

Heating Safety testing in swine

In the MRI suite, a large (45 Kg) swine was instrumented with fiber-optic probes underneath all surface patches. Two MRI-compatible catheters were navigated into the left atrium and right atrium, respectively, and then left in maximal off-(magnetic)-center positions. FSE imaging at 4 W/kg specific absorption rate (SAR) was performed for a continuous duration of 45 minutes, followed by a balanced steady state free precession (bSSFP) sequence of the same duration and SAR. The temperature below the patches was continuously recorded. We did not measure the temperature at the catheter electrodes. The animal was then sacrificed and the skin below the patches, and the atrial walls, were examined for burns.

Swine Navigational Experiments

In five swine navigational experiments, the same procedure was followed. In the animal lab, a VDT-only MR-compatible EP catheter was inserted from the jugular vein into the coronary sinus (CS) under X-ray guidance. In addition, a 3.3mm S2-tipped 70 cm- long unbraided sheath (SJM) was advanced from the femoral vein into the right atrium under X-ray guidance, and following trans-septal puncture, advanced into the LA. The CS catheter and the sheath were sutured, so that they would not move during the animal's transport to the MRI room.

After the animals were transferred onto the MRI table and connected to monitoring and ventilation, the animals were inserted into the MR scanner and a sagittal, ECG-gated gadolinium-contrast-enhanced (0.66mL/kg Magnevist™) 3D MR Angiography (MRA) scan was performed during a forced breath-hold, covering the apex to the aortic arch. MRA parameters: TR/TE/θ= 2.9msec/0.8msec /25 degrees, 160×256, 28–30×28–30 cm FOV, 2.8mm slice width (interpolated to 1.4mm), ±83.3 kHz bandwidth, end-systolic ECG-triggered, 200 msec imaging window/cardiac-cycle, 32–40 slices/slab, 1 average.

Data from this scan was segmented on the VDT workstation using EnSite™ Verismo™ Segmentation software, providing an MRI-based navigational roadmap.

After the MRA scan, two additional EP catheters were inserted; into the femoral arterial and venous ports, and thereafter navigation in the left and right heart was performed. Initially, the cardiac anatomy was mapped using the VDT System by sweeping through the anatomy, creating a VDT-based roadmap. The segmented MRI-based roadmap was then registered with the VDT-based roadmap using EnSite™ Fusion™ Registration software⁴⁹, which has a positional registration error of 2 to 7 mm, so that navigation could later be performed using either the MRI-based roadmap or the VDT-based roadmap. These two roadmaps were accurately co-registered only in a small area, since the VDT coordinate frame was non-linearly distorted relative to the MRI-based coordinate frame. Although this discrepancy could have been reduced with manual registrations of the VDT map at multiple anatomic positions, manual registration was not performed.

With the VDT-based roadmap complete, navigation was performed in the left side of the heart, left atrium and ventricle, using the venous approaches, utilizing the sheath in the left atrium, and also via arterial (retrograde”) approaches (femoral artery-aorta-left ventricle). Navigation in the right side of the heart was also performed, from the inferior vena cava, into the right atrium and into the right ventricle or superior vena cava. Locations and wall-voltages were continuously acquired.

To estimate the effects of the static magnetic field on VDT locations, the 3 EP catheters were left in fixed locations, their VDT position was recorded, and the animals were then pulled out of the magnet bore. The VDT position of the catheters outside the bore was recorded and navigation was then resumed on the MRI table without coordinate re-registration.

To test the effect of the VDT system on MRI image quality (IQ), ECG-gated high-resolution cine bSSFP images were acquired during forced breath-holds. 2D bSSFP parameters; TR/TE/θ=4 msec/2 msec/30 degrees, 320×256, 30×30 cm FOV, 5 mm slice width, ±64 kHz bandwidth, 2 averages, 20 reconstructed cardiac frames, 15–18 sec/slice.

To test the sensitivity of the VDT data to MR sequences, the catheters were placed at fixed anatomic positions with the animals inside the bore and VDT locations were recorded. MR imaging was then performed using several sequences, and changes in VDT location during imaging were recorded. The MR pulse sequences tested were: A) 2D spoiled Gradient-echo (GRE), TR/TE/θ=4–500 msec/1.6–5 msec/45 degrees, 256×128, 35×35 cm FOV, 3–8 mm slice width, ±16–125 kHz bandwidth, 1 average. B) 2D Fast Spin Echo (FSE), TR/TE/

$\Theta=10\text{--}500\text{ msec}/5\text{--}40\text{ msec}$, Echo-Train-Length=2–16, 256×128 , $35\times35\text{ cm}$ FOV, 5–8 mm slice width, $\pm16\text{--}125\text{ kHz}$ bandwidth, 1 average. C) 2D bSSFP, TR/TE/ $\Theta=3\text{--}5\text{ msec}/1.5\text{--}2.5\text{ msec}/45\text{ degrees}$, 256×128 , $35\times35\text{ cm}$ FOV, 5 mm slice width, $\pm64\text{--}125\text{ kHz}$ bandwidth, 1 average. No ECG gating was used. I

In all cases, the pulse sequences utilized were commercial sequences. The GRE and bSSFP sequences employed gradient trapezoids at the scanner's maximum slew rate of $150\text{ T m}^{-1}\text{s}^{-1}$, with amplitudes of up to 40 mT m^{-1} . No attempt was made to reduce slew rate for the purposes of reducing gradient noise.

For each sequence, MRI IQ was compared between imaging with and without simultaneous VDT to assess the possible effects of the VDT system.

Results

Phantom Experiments performed without the hardware blocking circuit

Figure 4D shows a LA map displayed on the Velocity workstation using dual-VDT and MR-tracked catheters, showing a catheter navigated into one of the pulmonary veins. MRI-tracking views taken with the MR-tracking system³⁸ show MRI-tracking renditions of this catheter, overlaid on real-time MRI images (Fig 4E, 4F) and on a pre-acquired 3D GRE slice (Figure 4G). Figure 4H shows the VDT display of two catheters simultaneously navigating through the phantom. Simultaneous VDT during MRI-tracking acquisitions at both 10 and 20 fps succeeded without any visible reduction in the MRI-tracking SNR. VDT signal averaging was required, reducing the tracking rate from 90 to 12 fps to maintain VDT accuracy. Low-bandwidth ($\pm32\text{ kHz}$) 2D MRI acquisitions were performed simultaneously with VDT with $<2\%$ degradation of MRI SNR, allowing VDT at 12 fps, while high-bandwidth MRI acquisitions ($\pm128\text{ kHz}$) prevented VDT acquisition, although the MR image SNR (Fig. 4D, 4E) was not noticeably disturbed. Correct voltage device tracking was restored 11 msec (one VDT acquisition cycle) after MRI imaging was suspended. The primary cause of the VDT artifacts encountered in these experiments (Fig. 2D) was saturation of the VDT amplifiers due to induced voltages from the MRI gradient ramps and the RF-blanking pulses.

These experiments proved that it was possible to perform VDT simultaneously with MRI imaging without sacrificing MRI SNR. They also revealed high-slew-rate MRI sequence interference with VDT, which led to construction of artifact-blocking circuits.

Swine testing of lead heating

Temperature measurements performed below the swine VDT patches showed an increase of temperature of $<1.5^\circ\text{C}$ over the entire duration of the FSE and bSSFP scans. When the patches were removed, no burns were detected. These scans also established the effectiveness of surface patch slotting in localizing bSSFP eddy-current-related image artifacts. With slotted patches, artifacts were $<20\text{ mm}$ in depth, with a horizontal extent equal to the patch areas. Smaller artifacts were observed with FSE and GRE. In addition, slotted patches resulted in $<5\%$ volumetric SNR loss, as compared to un-slotted patches which reduced SNR by 15–20%, relative to imaging without the VDT system.

Swine Navigational Experiments

Figure 5 shows navigational experiments performed inside the MR magnet in two swine. Segmented MRA-based maps (Figure 5A), were registered with the VDT-based EAM maps, (Figure 5B) and both were displayed on the EAM workstation. Three MRI-compatible VDT catheters are shown during navigation in the right and left heart. These catheters were manipulated by two electro-physiologists standing in front of and behind the magnet, viewing real-time maps on MRI-compatible in-room monitors²⁸. The VDT-based anatomy was geometrically distorted, so the MRI-based map assisted navigation.

Views of two tracked catheters during navigation inside a swine RSPV (Figure 5C1–C3), show one catheter maintained inside the swine's CS, recording heart motion due to respiration and partially (superior-inferior motion only) correcting for it, while the second catheter mapped the pulmonary vein and then navigated within it. After positioning the catheter and marking the point with a shaded contour (Figure 5C2), the swine was moved out of the magnet, so that the foreground catheter image showed the change in the CS catheter's VDT location between its position inside and outside the MRI. Changes were <2 mm, both along the catheter axis and perpendicular to it.

High-resolution bSSFP-cine scans acquired with the VDT system On are shown in Figures 6A–C. Compared to images acquired with the VDT system Off, the SNR difference was <3%. The catheter tips appear as bright spots, since the larger metallic tip-electrodes increased the effective RF flip angle in their proximity and produced magnetic-susceptibility variations. Despite the increase in effective flip angle, no burns were detected in the swine hearts in post-mortem histology.

The effect of MRI imaging on VDT location and EGM is shown in Figure 7. Catheters were navigated to specific anatomic regions before imaging was performed, and the change in location and EGM was observed after imaging was initiated. In Figures 7A and 7B, a TR=67msec GRE sequence was used and the blocking circuit was utilized. Catheter position (Figure 7A) remained stable during the entire imaging session. Figure 7B compares EGM appearance without (blue-blue-blue trace) and with the blocking circuit (blue-red-blue trace). The circuit had no affect without imaging, and it reduced the degree of noise during imaging. Comparing EGMs acquired without and with imaging, it is seen that the blocking circuit reduced EGM amplitude according to the fractional time within the TR during which it was On. Automatic correction for this amplitude change is planned in future work. Figures 7C,7D and table 1 show the VDT location distribution for four VDT electrodes, in two catheters, over 10 sec increments, without and during imaging. The median electrode location changed by 0.13 to 0.37 mm, while the dispersion (median absolute deviation) increased by 0.05 to 0.2mm during the scan. Since some of the positional standard deviation was due to respiratory or cardiac motion, this slight increase in positional oscillation was hard to visually detect.

Suppression of MRI-based noise affecting VDT location succeeded in TR>32 msec GRE and FSE sequences. In very short TR sequences, such as TR=4msec TR bSSFP sequence (Figures 7E and F), the MRI-based location noise was insufficiently suppressed, so that catheter locations do not appear (Figure 7E), even with the blocking circuit, while EGM data

from both catheters was reliably recorded (Figure 7F). After imaging was stopped, VDT localization was restored in 30 seconds. Multislice TR=67msec GRE MRI images (Figure 7G) are shown, highlighting the position of the three catheters. GRE IQ was in-significantly (<5% of SNR) affected by the VDT system.

Supplementary Movie #1 shows the effects of a TR=32 msec FSE scan on VDT location and EGMs inside the MRI. Most of the motion observed is probably due to respiration. Supplementary Movie #2 shows simultaneous manipulation of two catheters, with the swine outside the magnet, with continued positional tracking. There was no need to re-register or re-acquire the navigational roadmaps after the swine was moved out of the magnet.

Dominant differences in the EGM between that seen outside the magnet and inside it are due to the MHD effect, which superimposed a voltage peak during the cardiac-cycle's S-T segment. This effect, commonly observed on surface ECGs acquired inside the MRI⁴⁶ is primarily due to flow in the aortic arch. By navigating a catheter into the aortic arch and deflecting the tip so that it lied perpendicular to the flow direction (Figure 8A, upper trace), it was possible to maximize this effect, demonstrating an S-T peak larger than the QRS complex, whereas in the descending aorta (Figure 8A, lower trace, and Figure 8B, upper and lower traces) it was no longer dominant, but still significant.

Discussion

VDT of EP catheters inside an MR scanner was demonstrated, along with acquisition of MR images without significant SNR reduction. It was also possible, using the blocking circuit, to perform MR-tracking simultaneously with VDT, as well as to acquire TR>32 ms MRI-images simultaneously with VDT. During imaging with TR<32 ms sequences EGMs were recorded with reasonable fidelity but location data was corrupted.

MRI induced noise in VDT catheter location were also reduced by employing software that ignored corrupted data, detected as a large signal input at the VDT amplifier. This algorithm was designed to remove MediGuide™ Technology induced artifacts⁴⁷, so that it imposed certain limitations on the TR of MRI sequences. Tailoring this method for MRI could reduce its current limitations.

The inability to successfully perform VDT during short TR MR imaging sequences was a result of the high duty cycle of the gradient-ramps which severely limited the data acquisition window for VDT. Minimizing this duty cycle limit is a subject of current work, but even with existing technology, it is still possible to acquire short TR sequences, by interleaving in time between catheter manipulation and imaging. For example, if imaging of the surroundings of a catheter tip during navigation is desired, one could track for a few seconds to determine the catheter-tip coordinates, image for a few seconds using the determined imaging location/orientation, and then track again for a few seconds. Using such an interleaved imaging-and-tracking protocol, the effect on the interventional workflow could be minimal.

A method to reduce the current sequence limitations includes the use of “quiet scans”, sequences offered by several manufacturers for imaging of sensitive patient populations

such as non-anesthetized pediatric patients. These sequences utilize smaller gradient slew rates, and as a result will produce smaller amplitude (electric field) noise^{46,47}.

A possible method to reduce the short TR limitation would be to calculate, or measure, the electric field (noise) produced by each gradient waveform^{46–47} within a given pulse sequence and thereafter to only block those gradient patterns that produce noise higher than the VDT amplifier's saturation voltage (100 mV). Such an approach may allow sufficient acquisition time within a TR<4ms pulse sequence for EGM and LOC signal acquisition, allowing for further EGM signal “cleaning” using signal processing.

A limitation of this study is the incomplete assessment of the risk of heating, since we did not measure the temperature rise within the catheters themselves. Clinical-grade VDT catheters should include filters to prevent the propagation of radio-frequency waves on all the transmission lines. This should be easier to achieve with VDT-tracked catheters, since VDT signals are in the low-energy Hz to kHz range, and RF propagation is not required, so that simple low-pass filters may suffice. This should be compared to active MRI-tracking, where differential mode RF propagation must be maintained³⁸.

The ability to perform VDT inside the MRI enables the continuous tracking of catheters outside the MRI, inside the MRI, as well as during transitional stages. This capability can increase the efficiency of EP procedures which require the movement of the patient from the conventional EP suite into the MR scanner. Leaving a catheter inside the coronary sinus, for example, allows detection of the heart's movement, if it occurs at any point during the procedure, and also facilitates re-registration of image data to the original frame of reference. This ability to detect motion and to correct for it can have significant consequences on the monitoring of radio-frequency therapy delivery, and thereby on procedural outcomes and duration.

A practical example is our current practice, which is typical of sites where MRI information is utilized to improve EP procedures. In our X-MR suite⁵⁰, atrial fibrillation radio-frequency ablations are performed on patients in which AF recurred after a first ablation. This procedure starts in the MRI, where MR Angiography (MRA) and Late Gadolinium Enhancement (LGE) map LA luminal anatomy and map the scarred wall, respectively. The scar map provides the location of gaps that caused the AF recurrence. The patient is then transferred to the EP room, where catheters are inserted and brought to the LA. Registration between the MRI and VDT anatomy is now required, which requires touching several cardiac landmarks that can be identified on both VDT and MRI. EP to MRI transfer and the registration requires 45–60 minutes. Once registration is completed, the scar gaps can be mapped electrically and ablated. Once ablation is complete, all catheters are withdrawn and the patient is moved back into the MRI. This process requires another 45–60 minutes. In the MRI room, MRA, LGE and T2-weighted imaging are performed, assessing the scar and edema created by the current procedure, and attempting to determine whether the current lesions will remain permanently. Once MRI imaging is complete, the patient may be moved back to the EP suite for further ablation, which requires reinsertion of catheters and a second registration, which takes 45–60 minutes, before “touch-up” ablation can be performed. In total, 135–165 minutes of overhead are added to the EP procedure, which significantly adds

risk, since the patient is intubated during this entire period. Leaving a VDT tracked catheter in the coronary sinus might remove 90–120 minutes of overhead, and it would increase ablation accuracy, due to a reduced need for re-registration.

In addition to improving current practice, an MRI compatible VDT-system opens the door to using multiple devices which can be tracked using VDT. These include interventional devices such as sheathes and balloons. Since additional imaging and therapeutic devices can be easily outfitted with electrodes, these devices can currently be used in the combined X-MRI suites with co-registration, and brought into the MRI bore once they are MRI-safe. For example, non-magnetic ICE catheters could be brought into the MRI bore, and used during periods when the MRI is not pulsing.

Catheters capable of both VDT and MR-tracking provide greater benefits than MR-conditional VDT catheters alone, since they enable accurate co-registration of VDT-based anatomic maps with MR-based maps throughout the study volume. Such co-registration utilizes the principle that two MR-tracked points on a catheter precisely scale the relative distances measured between electrodes on this catheter. Such scaling can remove the geometric distortion currently present in VDT maps.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. Voltage-Based Device Tracking principles. (A) Three pairs of electrodes: Superior/ Inferior, Right/ Left, Head/ Feet, and a reference electrode (SysRef) are placed on the patient skin. (B) When a voltage (10 mV) is placed between the Right and Left electrodes, a voltage potential is set up between this electrode-pair (numbered vertical dotted lines). If a catheter is placed within this potential field, measurement of the voltage drop between these electrodes and the Left electrode (7 and 3 mV, arrows) provides the relative position of these electrodes in the Right-Left direction. Similar voltages in the Superior-Inferior and Front-Back directions provide localization in the other two directions, resulting in 3D positional localization in the VDT frame of reference.

Figure 2.

Voltage-based device tracking system in the MRI suite. (A) Block Diagram of the VDT system, showing transmission (left) and reception (right) units. Localization signals are transmitted to patch pairs on the patient surface. Each catheter electrode transmits both positional information (LOC, black) and intra-cardiac ECGs (EGMs, red) to the VDT amplifier. Highlighted Delay lines and Switches were added to permit data recording during MRI imaging, by removing induced noise during gradient ramps and RF-pulse transmission. (B) Positional localization signals (LOC) in the three coordinate axis (SI, LR, AP) received in a human in the absence of imaging, and (C) during an MRI gradient echo sequence. During imaging the amplifiers were saturated (C) due to strong induced noise. Saturation was partially alleviated (D) by blocking signal transmission to the amplifiers ("Blanking") during periods of extreme imaging-related noise.

Figure 3.

Modifications to the SJM VDT system for MRI-compatibility. (A) surface patches were slit at 1cm increments, and 100cm leads were replaced with 4cm leads. (B) Single-conductor transmission lines in the MRI room were replaced with 2mm-diameter coaxial cables. Ferrites were placed on the coaxial cables at half-wavelength increments to reduce RF propagation, and the shields were attached to the penetration panel via capacitive connectors (DC blocks). Outside the penetration panel, Low pass filters prevented RF noise from entering the MRI room. (C) Signal switching circuit, placed between the penetration panel and the VDT amplifiers, contains delay-lines/low-pass filters and SPDT switches which disconnect signals from amplifiers during noisy signal intervals. Noisy intervals are detected by a circuit (lower row) that receives gradient and RF-unblank waveforms from the MRI system cabinet. (D) Non-irrigated MRI-tracked and VDT-tracked EP catheters used in phantom studies. (E) Irrigated VDT-tracked EP catheters used in animal studies.

Figure 4.

MRI-compatible VDT phantom testing using dual VDT-tracked and MR-tracked catheters; (A) wet-lab showing 7 electrode connections (B) left atrial model used in the phantom studies, (C) wet-lab with inserted LA phantom, and filled with saline solution. (D) Insertion of dual-traced catheter into a PV under VDT system guidance, utilizing EAM mapping of the atrial model, with simultaneous MR-tracking display using MR-tracked micro-coil positions overlaid on two high-speed (high-bandwidth) MRI images (E, F). (G) High-

resolution 3D MRI images used were acquired as MR-tracking landmarks, with two VDT catheters tracked at 12 fps (H) (yellow arrows) during the MRI imaging.

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Figure 5.

Navigating inside MRI in swine cardiovascular anatomy using VDT. (A) Segmented MRA map of swine anatomy, overlaid with positions of 3 catheters (dotted ellipses; in RA, RV, CS). LV: left ventricle, LA: left atrium, RV: right ventricle, RA: right atrium, SVC: superior vena cava, CS: coronary sinus, RSPV: right superior pulmonary vein] (B) Map produced using VDT system by touching vessel-/chamber-walls using the 3 catheters. Maps (A) and (B) were registered on the SJM console, which allowed following VDT catheter positions during navigation on both maps. (C1–C3) RSPV navigation in another swine. Yellow

catheter in coronary sinus, white catheter is in RSPV. CS catheter in C2 was tracked to detect positional changes between its position inside the MRI (shaded contour) and outside (foreground contour) the scanner.

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Figure 6.
(A–C) Swine high resolution bSSFP cine images acquired with VDT-tracked catheters in place.

Figure 7.

VDT Navigation during MRI imaging. During TR=67msec GRE scan; (A) location map with blanking enabled shows stable catheters (B) Intra-cardiac ECG (EGM) traces without (Blue trace) and with (Blue-red-Blue trace) blanking enabled, acquired during periods during and without MRI imaging. Note smaller EGM gradient artifact with blanking enabled. (C) Frequencies of location of distal electrode and proximal electrode in VDT-catheter #1 over a 10 sec period *without imaging only* (blue), *both without and during imaging* (red), versus *only during imaging* (orange). (D) same as (C), but for VDT-catheter

#2. During Non-gated TR=4 msec bSSFP scan; (E) location map, with blanking enabled, doesn't show catheters (F) two EGM traces, taken with blanking enabled, show that proper EGMs were recorded during the imaging period. (G) TR=67ms GRE images acquired with the catheters in place.

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Figure 8.

Magneto-hydrodynamics at its source. During navigation in the left side of the swine heart, very strong MHD peaks (white dashed circles) were seen during the ST segment of the cardiac cycle in EGM traces obtained in the (A) aortic arch (green arrow), considered the major MHD source. Weaker MHD signal was observed in imaging of the left ventricle (A, yellow arrow) and even in the descending aorta (B). These features were absent outside the MRI.

Table 1

Effect of MRI imaging with TR=67msec GRE on swine VDT-catheter electrode locations. Location of four VDT electrodes (Distal, Ring 1, Ring 2, Ring 3, Ring 4), in two catheters (#1, #2), was recorded over a 10-second interval without and thereafter during imaging. Some of results are also shown graphically in Figures 7C and D.

Catheter	VDT Electrode	Difference in mean location Imaging Off vs. On (mm)	95% confidence interval differences Imaging Off vs. On (mm)
#1	Distal	0.33	0.33 – 0.37
	Ring 2	0.37	0.33 – 0.40
	Ring 3	0.37	0.33 – 0.40
	Ring 4	0.32	0.30 – 0.35
#2	Distal	0.18	0.15 – 0.22
	Ring 2	0.13	0.10 – 0.17
	Ring 3	0.14	0.10 – 0.17
	Ring 4	0.21	0.17 – 0.24