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MR-guided Passive Catheter Tracking for Endovascular Therapy

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Synopsis

The use of MR guidance for endovascular intervention is appealing due to its lack of ionizing radiation, high-contrast visualization of vessel walls and adjacent soft tissues, multiplanar capabilities, and potential to incorporate functional information such as flow, fluid dynamics, perfusion, or cardiac motion. Concurrent advances in the design of “real-time” MR fluoroscopy pulse sequences and the development of MR compatible endovascular devices have facilitated progress from initial *in vitro* and animal feasibility studies of the initial use of MR guidance for endovascular interventions to human clinical applications. This review highlights current state-of-the-art imaging techniques and hardware used for passive tracking of endovascular devices in interventional MRI, including negative contrast, passive contrast, non-proton multispectral, and direct current techniques. The advantages and disadvantages of passive tracking relative to active tracking are also summarized.

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Keywords

Endovascular intervention; MRI; passive tracking; device tracking; MR angiography; Real-time MRI

Introduction

The use of MRI for guidance of endovascular interventions has emerged as a feasible alternative or adjunctive imaging modality to digital subtraction angiography (DSA). MRI guidance offers several potential advantages. The absence of ionizing radiation during MRI is particularly attractive, as substantial radiation exposures to both the patient and the interventionalist may be received using DSA[1, 2]. In addition, although DSA is limited to luminal information, MRI can also provide high-contrast visualization of the vessel wall and adjacent soft tissues with moderately high spatial resolution, allowing for the identification of vulnerable atherosclerotic plaques, the ability to monitor the effect of endovascular therapy on adjacent tissues and to recognize possible complications, such as vascular perforation, hemorrhage, and infarction, all in real-time. MRI also offers unrestricted multiplanar imaging capabilities and the opportunity to obtain physiological and functional information such as flow velocity, 4D flow imaging and computational fluid dynamics, temperature, diffusion, and perfusion.

Concurrent advances in the design of “real-time” MR fluoroscopy and MR angiography pulse sequences, and the development of MR safe and compatible endovascular devices have facilitated progress from initial *in vitro* and animal feasibility studies of MR guidance for endovascular interventions to their initial translation into clinical care. This review highlights current state-of-the-art imaging techniques and hardware used for passive tracking of devices in endovascular interventional MRI.

The vascular interventional MRI suite

The use of MRI guidance for endovascular intervention requires balancing multiple trade-offs involving image quality, spatial and temporal resolution, patient accessibility, field-of-view, and cost. Higher field strength cylindrical bore scanners are capable of better image quality; however, they require relatively small diameter bores compared to open MRI scanners and therefore limit patient accessibility. Currently, open bore MRI scanners do not provide adequate field strength, gradient strength, or field uniformity for endovascular intervention[3]. Clamshell and double-doughnut shaped bores were initially introduced to improve patient accessibility during iMRI. Fortunately, arterial access at the groin for most endovascular procedures permits longer distances (40-80 cm) between the interventionist standing at the opening of the magnet bore and the target tissue of interest in the head, neck chest, or abdomen, making commonly available cylindrical (or “closed”) bore clinical MR scanners suitable. Cylindrical bore magnets with shorter bore lengths (125 cm instead of 160 cm) and/or with larger bore diameters (70 cm instead of 60 cm) have been developed, allowing easier patient access and are commercially available. Most research sites now have hybrid systems consisting of a short-bore cylindrical MRI and DSA unit connected by a single “floating” table, also known as XMR systems (Figure 1, Movie 1). This allows for

use of either imaging modality for different parts of a procedure and permits the use of DSA back-up in case of difficulties during MR guidance.

Real-time MRI and MRA techniques for endovascular interventional MRI

Image acquisition during MR guided endovascular procedures must be rapid enough to allow the interventionist to visualize changes within the patient as devices are externally manipulated in real time Several pulse sequence approaches to achieve higher temporal resolution MR imaging for MR-guided interventions -- also known as “real-time” MRI or MR fluorosocopy -- have been explored. Higher temporal resolution usually comes at the expense of spatial resolution. Nevertheless, MR fluoroscopy sequences used today allow sufficient spatial and temporal resolution to track endovascular devices. Since each pulse sequence offers unique contrast characteristics, the ideal sequence will depend on the properties of the device being tracked and the passive tracking method being utilized for device tracking (see passive tracking section below), the contrast properties of the tissue of interest, and the need for intravascular contrast. Commonly used MR fluoroscopy sequences are summarized in Table 1.

Steady state free precession (SSFP) sequences have become the preferred sequences for endovascular interventional MRI. SSFP provides excellent bright blood contrast and visualization of devices employing passive tracking due to the high temporal resolution (short TR), high SNR, and greater sensitivity to magnetic field inhomogeneities[4-13] (Figure 2).

T2-weighted turbo spin echo (TSE), rapid acquisition with relaxation enhancement (RARE, HASTE) sequences may also be used for verification of device location due the flexibility in image contrast and relatively fast acquisition of the pulse sequence. Additional methods to decrease image acquisition time include the use of parallel imaging[14], key-hole imaging[15], alternative k-space sampling such as spiral or radial trajectories, asymmetric truncated k-space sampling strategies[16], and more recently, the exploitation of k-space data sparsity using compressed sensing[17]. Coupled with high-performance multi-processor computers, real-time MR-guidance is now capable of up to 10 frames/second, which is comparable to the temporal resolution of x-ray fluoroscopic guidance for most clinical endovascular interventions (15-30 frames/second).

MR angiograms with intraarterial gadolinium contrast may also be used for MR guidance of endovascular procedures[18] (Figure 3). Similar to DSA, contrast-enhanced MRAs can be overlaid onto subsequent image acquisitions for vascular road-map guidance[19-21]. Although intra-arterial injection of gadolinium chelate is an off-label use, no adverse events have been reported so far. The U.S. Food and Drug Administration (FDA) total dose limit for gadolinium chelate is 0.3 mmol/l/kg body weight. A low-dose gadolinium injection protocol requires[18, 22] taking into account the blood flow rate in the artery of interest and avoids excess concentrations of gadolinium (resulting in T2/T2* spin-dephasing effects) or inadequate gadolinium concentrations (resulting in lack of T1 shortening) [20, 24]. Since contrast delivery is local rather than systemic, much lower doses are needed, and thus, repeat intra-arterial contrast administration is practical. Gadolinium contrast dose limit issues may

be at least partially ameliorated by using new gadolinium-based contrast agents that remain in the blood-pool for longer periods of time and provide prolonged vascular enhancement (e.g., Gadomer-17, Vasovist)[23, 24].

Different MRI pulse sequences may be interleaved and swapped as the procedure demands as the ideal pulse sequence for device visualization or tracking often differs from that needed for vascular roadmapping, 2D and 3D visualization of tissue near the device, or therapeutic monitoring. Endovascular interventional MRI requires real-time changes in scan plane and pulse sequence parameters and MRI vendors and third party providers have started to develop interactive display and MRI control software that allow adaptation of parameters (e.g., slice orientation, slab thickness, FOV, or contrast) (Figure 4).

Tracking of endovascular devices

Accurate, fast, and reliable visualization and localization of endovascular devices are essential requirements for safe and successful endovascular procedures. In DSA, catheters, guidewires, stents, clips, and pacemaker leads are visible due to the increased x-ray absorption of high atomic number metal components and markers relative to blood and tissues. The strong magnetic field in the MRI environment, however, restricts the use of ferromagnetic metals commonly used in the manufacture of guidewires, stents, and in the braiding incorporated within catheter walls for improved torquability and pushability. In addition, the plastic polymers commonly used in catheter construction are very difficult to visualize on MR imaging. The increasing availability and clinical utility of MRI scanners has created a significant demand for MR compatible implants and devices such as titanium aneurysm clips and MR compatible pacemakers. MR compatible devices for endovascular interventional procedures, in contrast, are lacking due to a lower demand, high costs, paucity of MRI scanners dedicated to image guided procedures, relative familiarity with alternate modalities for image-guidance of procedures, and lack of training in MR-guided intervention. Nevertheless, several investigators have developed methods for endovascular device localization and visualization, or tracking, in the interventional MRI environment. These methods of device tracking can be divided into two main categories: active and passive.

Passive tracking methods

With passive tracking, endovascular devices can be tracked on MRI by incorporating markers or materials in their construction that enhance, reduce, or distort the T1 and/or T2-weighted MR signal relative to blood or tissues, either focally using a marker(s) at the tip, or along the entire length of the device. Passive tracking methods are summarized in Table 2.

Negative contrast

Passive tracking using negative contrast uses marker materials in device construction that cause localized magnetic field inhomogeneities resulting in dephasing of adjacent proton spins and a local signal void or susceptibility artifact on MRI without the use of active RF components[9, 25-32] (Figure 2, Table 2).

The marker materials may be ferromagnetic (e.g., stainless steel type AISI 410, nickel), ferrimagnetic (e.g., copper zinc ferrite), or strongly paramagnetic (e.g., dysprosium oxide). The focal susceptibility artifact associated with this kind of marker is most easily visualized on T2*-weighted MR imaging sequences. To increase the contrast between a passively tracked MR compatible catheter or guidewire and background tissues, subtraction of a baseline tracking image can be used for background tissue suppression, and can be overlaid onto a previously acquired MR angiogram for road-mapping[33-35]. An adaptive subtraction technique that automatically selects the most suitable reference image from a dynamic series for subtraction resulting in reduction in degradation of imaging by patient motion has also been described[35].

During negative contrast passive device tracking, reliable device visualization depends on contrast-to-noise ratio (CNR) and the visibility of the susceptibility artifact in the subtraction image. The CNR in the tracking images depends on the susceptibility artifact induced by the marker and the SNR of the tracking image[31]. The SNR can be controlled by sequence design, the size and amount of susceptibility artifact by the magnetic moment (*m*) of the marker material, and the echo time[36]. For diamagnetic and paramagnetic materials, *m* is proportional to the main magnetic field of the MR scanner. For ferromagnetic and ferrimagnetic materials, a nonlinear relationship exists, meaning the size of susceptibility artifact is very similar through a range of field strengths[31]. As a result, with ferromagnetic and ferrimagnetic markers, in order to achieve the same size of artifact, less material is required than with paramagnetic markers[31]. If the entire catheter is doped with the negative contrast material, the artifact visibility depends on field strength and device orientation to B_0 [25, 26, 37]. The artifact i

Similar to diagnostic MRI, susceptibility artifact size is also dependent on the imaging sequence used. GRE sequences, for example, result in larger susceptibility artifacts and consequently poorer information on surrounding tissues, whereas spin-echo sequences produce smaller areas of signal cancellation[38]. The relative importance of loss of the anatomical information around the artifact on MRI will vary with different procedures.

The signal void created by negative contrast markers can also be converted to a bright spot, or “white marker” (Figure 5). The white marker phenomenon can be elicited by dephasing the background signal with application of a slice gradient, while the signal near the marker is conserved because a dipole field induced by the marker compensates the dephasing gradient[39]. The MRI appearance of the white marker is determined by the marker magnetic susceptibility, echo-time, slice thickness, and gradient strength[39]. A white marker phenomenon for negative contrast markers can also be achieved using off-resonance techniques such as low flip angle SSFP sequences (or FLAPS)[32, 40]. The intensity and the spatial extent of the off-resonance contrast are strongly dependent on TR, flip angle, field strength and T2/T1 of the medium. Although off-resonance techniques may result in increased accuracy of the location of the marker, this technique does not eliminate the potential for loss of signal from adjacent anatomical structures due to the local magnetic field inhomogeneities caused by the markers.

A distinct feature of most passive tracking techniques is that the tracking mechanism cannot be switched on or off since the markers are built into the catheter. Recently, however, it was

shown that a passive marker using negative contrast -- in this case, a titanium cylinder built into a catheter tip -- could be mechanically switched “on” or “off” by sliding a graphite cylinder with a layer thickness calculated to cancel out the susceptibility artifact of the titanium marker[41]. However, this setup did not allow for use of the inner lumen of a catheter and the outer diameter of the device measured a relatively large 9 French.

Negative contrast passive tracking has also been shown using CO_2, either injected into a catheter after gadolinium contrast-enhanced T1-weighted MRA[23] or using balloon catheters inflated with CO_2[10].

Positive contrast

Passive tracking can also be achieved with “positive contrast,” using paramagnetic materials that cause focal T1 shortening (bright spot), such as gadolinium, within the lumen[9, 30] **or on the surface of a catheter**[42, 43] (Table 2, Figure 6). This is best visualized on a T1-weighted pulse sequence. The main drawbacks of passive tracking with positive contrast are the inability to use the lumen of contrast-filled single-lumen catheters or need for larger diameter multilumen catheters. Furthermore, positive contrast materials used for catheter coating can be relatively unstable. The T2* effect of gadolinium can also be exploited for positive contrast passive tracking. If the excitation frequency and bandwidth are properly chosen, specific concentrations of gadolinium contrast agent within a catheter can be excited without exciting fat or water spins[44]. The authors' term this technique “off-resonance contrast angiography”. In addition, complete background suppression can be obtained without image subtraction. This method is highly sensitive to catheter orientation within the B_0 field, however, which currently limits practical usefulness.

Non-proton multispectral contrast

Recently, passive tracking was demonstrated with catheters filled with contrast materials visible with non-proton magnetic resonance such as ^{19}F and hyperpolarized ^{13}C [45, 46]. Although these methods are capable of high CNR and SNR, the spatial resolution of this device tracking technique is lower, additional scanner hardware and software are required, and ^{1}H-MRI roadmap overlay is obligatory. In addition, due to the inability to use the lumen of the ^{19}F or hyperpolarized ^{13}C contrast-filled catheters, larger diameter multilumen catheters are necessary when using this technique as well.

Direct current

Another passive tracking method using negative contrast employs the local field distortions and resulting susceptibility artifact caused by application of small direct currents (10-150 mA) running through wire coils wound at a catheter tip (Figure 7)[47-49]. Device visualization using this method can be turned on or off using a DC current switch controlled by the interventionist and the visualization artifact can be adjusted by varying the amount of current applied. This method can also be combined with road-mapping with intravascular contrast[20] (Figure 3). With currents higher than those needed for catheter tip visualization, such coils may be used to steer a catheter tip into difficult turns[48, 49]. Despite the use of long wires running along the length of the catheter and resultant potential for RF heating, *in vivo*

testing has found endothelial damage from RF heating effects using this this method to be negligible[50]. With heat-dissipating saline flowing at 2 cc/s within a guide catheter in a larger vessel, no endothelial damage from ohmic heating of the wire or coil from current application for tip deflection was found at current levels under 300 mA[51]. The wire coils could also be used as RF receive coils, permitting concurrent active tracking or endovascular imaging[48].

Passive tracking techniques can also be combined with active tracking techniques and used simultaneously in the same procedure to track different devices. Omary *et al* actively tracked a 0.030-inch guidewire with a loopless antenna and passively tracked a 5-French catheter filled with dilute 4% gadopentate dimeglumine (positive contrast), allowing catheterization and stenting of swine renal arteries under MR guidance[52].

Advantages and disadvantages relative to active tracking

Active tracking techniques are beyond the scope of this article and discussed in detail separately in this issue. Advantages and disadvantages of passive tracking relative to active tracking are summarized in Table 3. The main advantage of passive catheter tracking over active tracking is simplicity as no additional MRI scanner software or electronics are required for the majority of passive tracking techniques. In addition, the absence of a wire connection from the device to the scanner for negative and positive contrast passive tracking eliminates the risk radiofrequency (RF) heating posed by long conductive wires needed for active tracking during MR image acquisition[53-55]. The decreased complexity of passive tracking also allows for better miniaturization of devices.

While a needle can be completely visualized on a single image acquired along the needle shaft, the course of an endovascular catheter within vessels is seldom in a straight line. Visualizing the catheter shaft, which in some cases can become redundant or looped during clinical catheterization procedures, may require prescription of multiple scan planes and placement of tracking markers proximal to the catheter tip. In DSA, the course of an endovascular catheter can be easily obtained using a single projection, or orthogonal bi-plane projections even within a tortuous vessel. Since previously acquired MRA images can be used as an anatomical road-map onto which catheter tracking can be overlayed, catheter guidance shares similarity to conventional DSA roadmap guidance. However, visualizing the course of a catheter is more difficult using MRI as the distal catheter may course out of the imaging plane(s) often unbeknownst to the interventionist during the procedure.

Since device coordinates are not registered with passive tracking methods and imaging plane is not automatically updated, automatic visualization and correct imaging plane selection is not possible with passive techniques. This is the principal disadvantage of passive tracking methods as the process of manually updating the imaging plane may be more time-consuming[56].

In a feasibility study of MR guided cardiac electrophysiologic ablation in humans, scan plane reorientation, when a passively tracked catheter fell out of plane, was usually achieved within one to three attempts. In some cases, however, it took up to 15 attempts, with up to 40 seconds until the catheter tip was relocated[57]. Thicker imaging slices are often used to

keep the passive marker within the imaging volume; however, increased slice thickness may decrease visibility of the passive marker and increases superimposition of anatomy. Furthermore, passive tracking methods may be hampered by the need for subtraction due to weak negative contrast of the passive markers to their background, especially if thick imaging slices are used. This subtraction leads to an undesired increased sensitivity to motion and flow artifacts[39]. The ability to automatically update the image plane with changes in position of a device using negative contrast susceptibility markers was recently shown, however, by interleaving a SSFP sequence with a "projection-reconstruction imaging with echo-dephasing" (PRIDE) sequence[58]. The latter sequence detects the focal area of susceptibility artifact due to the marker on an image along all three physical axes. After successful localization of the paramagnetic marker with PRIDE (with an absolute error of 4.5 mm), the location of the marker can be used to provide automated SSFP slice positioning using a dedicated real-time feedback link with update times comparable to existing active tracking methods[58] (Figure 8).

Clinical applications of endovascular device passive tracking

Despite a dearth of commercially available MR-compatible endovascular devices, the feasibility of commonly performed endovascular procedures using MR guidance has been demonstrated in numerous animal studies and early human clinical studies of a variety of vascular and cardiac procedures. Successful transjugular intrahepatic stent placement[59], coronary stent deployment[60], closure of atrial septal defects[61] and patent foramen ovale[62], aortic valve implantation[63], aortic coarctation repair[64], aortic dissection repair[65], renal angioplasty[52] and stenting[66], iliac artery stenting[6], placement and retrieval of IVC filters[67, 68], hepatic artery catheterization[69], recanalization of complete carotid occlusion[70], carotid stenting[71], aortic stent-grafting[72, 73], and hepatic arterial delivery of islet cells[74] have been successfully performed under MR guidance in animals. Compared to studies of non-vascular MR guided interventions, however, the lack of clinically approved endovascular devices for use in the MRI setting has limited the number of human studies. Endovascular procedures performed with MR guidance in humans include iliac[75], femoral and popliteal artery angioplasty[76], cardiac catheterization[77, 78], cardiac electrophysiological ablation[57] (Movie 2), and visualization of intramyocardial injection of gadolinium-DTPA[76].

However, the use of MR guidance for endovascular interventions has not yet been found to be superior, nor of additional benefit, compared to x-ray fluoroscopy/DSA guidance alone. MR guidance currently lacks sufficient spatial or temporal resolution for guiding meaningful clinical endovascular procedures[79], and the additional costs and procedure time currently favor DSA. Further development and testing of MR-compatible catheters, guidewires, stents, and other endovascular devices is needed, along with improvements in reliability of device visualization and tracking. Maintenance of a scan plane orientation that keeps the device tip in the field-of-view at all times remains challenging, especially with passive tracking techniques, but is obligatory for performing any endovascular intervention safely, quickly, and reliably. Future reports of improvements in clinical outcomes and the emergence of new therapies using MR guided endovascular therapies not possible under x-ray, DSA, or ultrasound guidance have the highest potential for driving further technical evolution of the field.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Abbreviations

DSA	Digital subtraction angiography
MRI	Magnetic resonance imaging
MRA	Magnetic resonance angiography
SSFP	steady state free precession
GRE	gradient recalled echo

Key Points

- Steady state free precession (SSFP) sequences have become the preferred sequences for endovascular interventional MRI.
- Passive tracking using negative contrast uses marker materials in device construction that cause localized magnetic field inhomogeneities resulting in dephasing of adjacent proton spins and a local signal void or susceptibility artifact on MRI without the use of active RF components.
- Passive tracking can also be achieved with “positive contrast” using paramagnetic materials that cause focal T1 shortening (bright spot), such as gadolinium, within the lumen or on the surface of a catheter.
- The main advantage of passive catheter tracking over active tracking is simplicity as no additional MRI scanner software or electronics are required for the majority of passive tracking techniques.
- Since device coordinates are not registered with passive tracking methods and imaging plane is not automatically updated, automatic visualization and correct imaging plane selection is not possible with passive techniques.

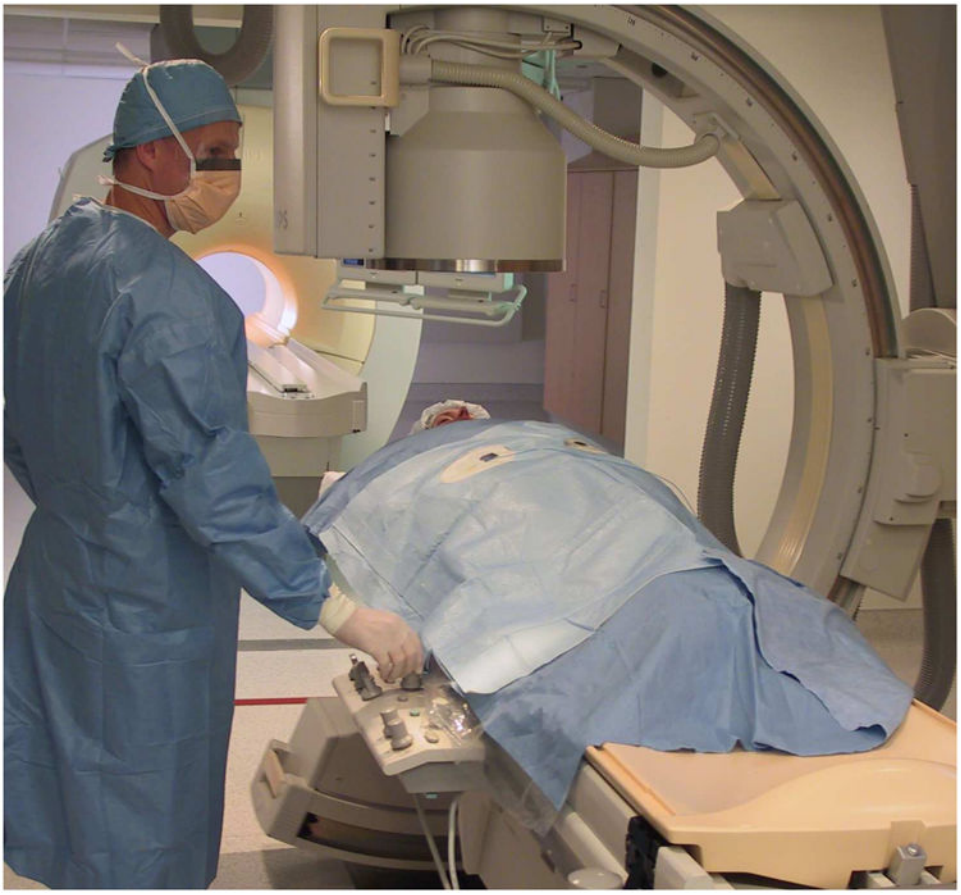

Figure 1.
Philips hybrid interventional XMR system at UCSF Medical Center, combining a clinical 1.5 T MRI scanner (background) and fully functional DSA unit (foreground), connected by a floating dockable table.

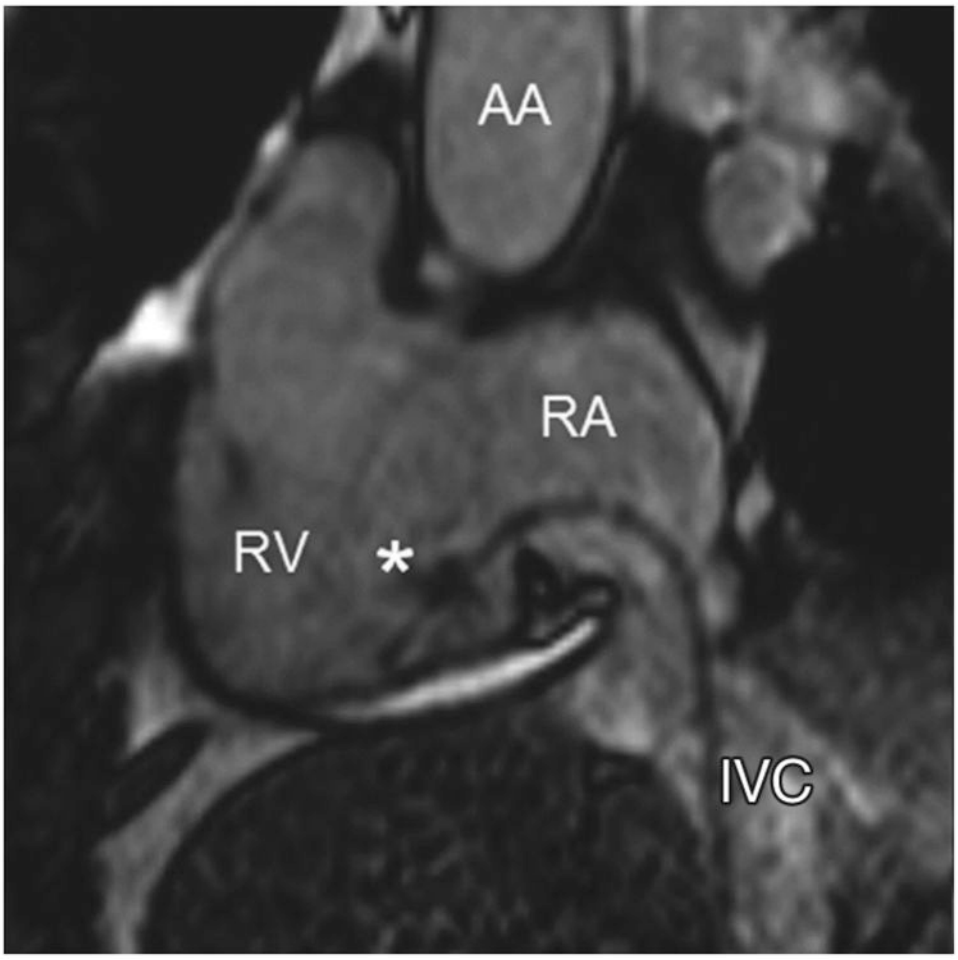

Figure 2.

MR images demonstrate passive catheter tracking within a human heart using a steady-state free precession sequence. The catheter tip is located above the cavotricuspid isthmus. *AA* = ascending aorta, *IVC* = inferior vena cava, *LA* = left atrium, *LV* = left ventricle, *RA* = right atrium, *RV* = right ventricle. * = passive marker near the catheter tip. (From Grothoff M, Piorkowski C, Eitel C, et al. MR imaging-guided electrophysiological ablation studies in humans with passive catheter tracking: initial results. Radiology 2014;271(3):695-702, with permission.)

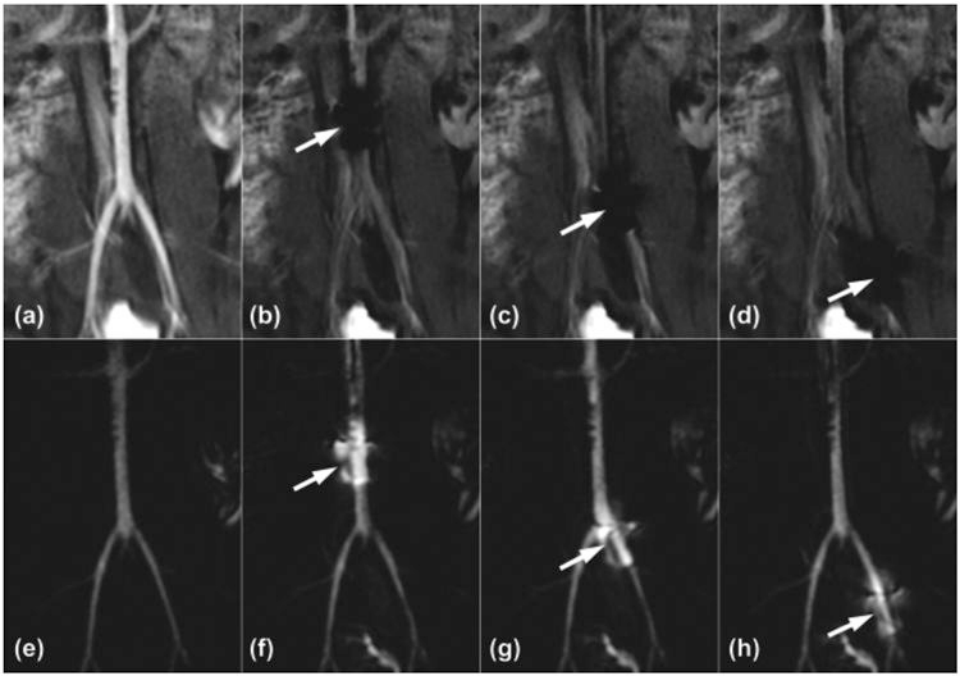

Figure 3.

The roadmapping approach is demonstrated in the distal aorta of a swine. The source images are shown on the top row and the roadmapping mode images are on the bottom row. An IA injection of Gd is initially used to highlight arterial anatomy (**A**) and establish the roadmap (**E**). The magnetically-assisted remote control catheter is then activated, producing a substantial signal void on the source images (arrows in **B–D**) and a signal enhancement pattern on the roadmap images that is superimposed on the arterial anatomy (arrows in **F–H**). The roadmap images can be used to track the device without losing visibility of the local arterial anatomy. (From Martin AJ, Lillaney P, Saeed M, et al. Digital subtraction MR angiography roadmapping for magnetic steerable catheter tracking. Journal of magnetic resonance imaging: JMRI 2014, with permission)

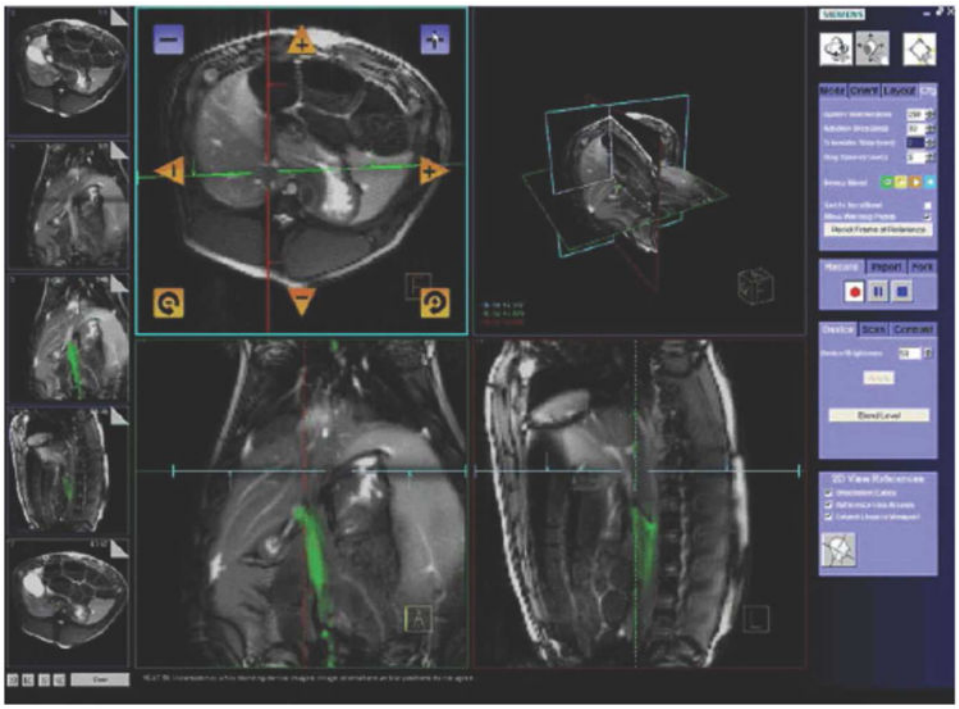

Figure 4.

Interactive Front End (Siemens) user interface for MRI-guided navigation. (From Saikus CE, Lederman RJ. Interventional cardiovascular magnetic resonance imaging: a new opportunity for image-guided interventions. JACC Cardiovascular imaging 2009;2(11): 1321-1331, with permission.)

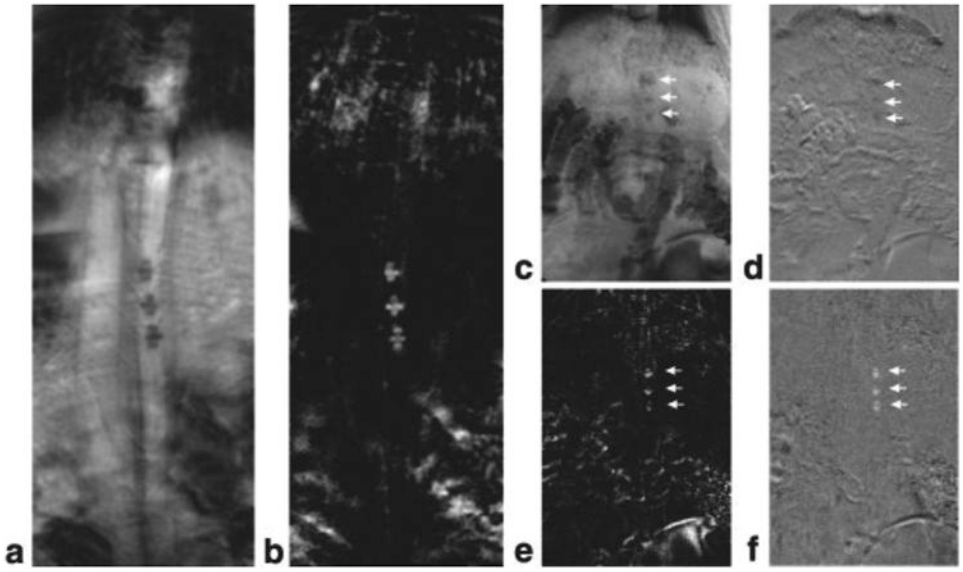

Figure 5.

Passive catheter tracking using "white marker phenomenon." **a,b:** In vivo imaging of three paramagnetic markers, mounted on a 5-F catheter, located in the abdominal aorta of a living pig, as visualized with (**a**) conventional gradient echo sequence (slice 30 mm, TE/TR = 4.6/60 ms, duration 22 sec) and (**b**) dephased positive contrast gradient echo imaging (white marker sequence with 1.9 cycles of phase across the slice) for similar acquisition parameters. **c–f:** Demonstration of performance of in vivo application of white marker tracking for a case with significant obscuring of the markers during in vivo tracking. For (**c**) unsubtracted and (**d**) subtracted conventional tracking, the markers are hardly seen, whereas the white marker tracking allows easy detection of the markers for both (**e**) unsubtracted and (**f**) subtracted positive contrast tracking. (From Seppenwoolde JH, Viergever MA, Bakker CJ. Passive tracking exploiting local signal conservation: the white marker phenomenon. Magnetic resonance in medicine: official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine 2003;50(4):784-790, with permission.)

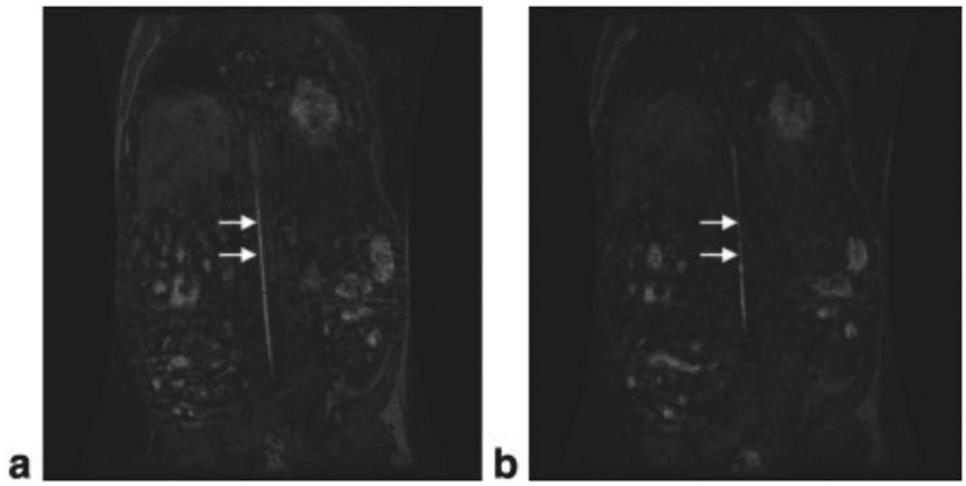

Figure 6.

Positive contrast tracking. Coronal T1 SPGR MIP images of a 6-French catheter filled with 4% Gd-DTPA (a) and a 4-French catheter coated using Design III (b) in a canine aorta. (From Unal O, Li J, Cheng W, et al. MR-visible coatings for endovascular device visualization. Journal of magnetic resonance imaging: JMRI 2006;23(5):763-769, with permission.)

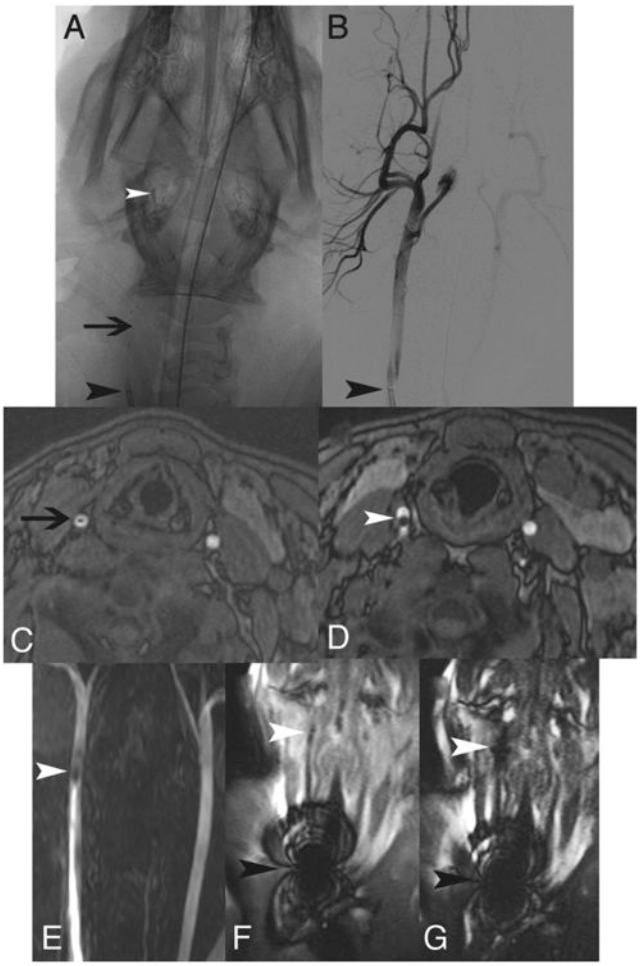

Figure 7.

Passive tracking using direct current coils. In vivo x-ray, DSA, axial MRA, coronal MRA MIP, and coronal SSFP. Unsubtracted x-ray image (*A*) demonstrates MARC catheter tip coils (*white arrowhead*), microcatheter shaft with lead wires (*black arrow*), and guiding catheter (*black arrowhead*) in the right CCA. Only the guiding catheter tip marker is readily evident on the equivalent DSA image (*B*). Susceptibility from the catheter shaft lead wires (*black arrow*) and catheter tip (*white arrowhead*) is seen on axial MRA (*C* and *D*), coronal MRA MIP (*E*), and coronal SSFP (*F*). With a 300-mA current applied (*G*), the catheter tip coils are more apparent (*white arrowhead*). Guide catheter tip artifacts resulting from a metallic marker band are very prominent on the SSFP sequence (*F* and *G*). (From Hetts SW, Saeed M, Martin AJ, et al. Endovascular catheter for magnetic navigation under MR imaging guidance: evaluation of safety in vivo at 1.5T. AJNR American journal of neuroradiology 2013;34(11):2083-2091, with permission.)

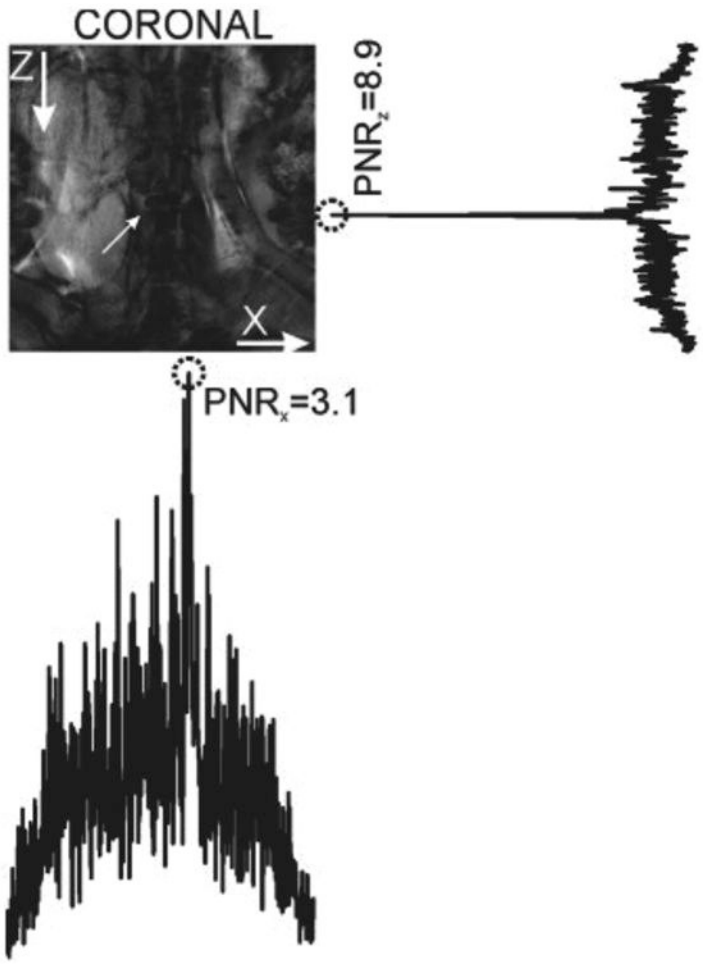

Figure 8.

Use of PRIDE for localization of a passively tracked negative contrast device in vivo: Real-time bSSFP coronal image displaying the placement of the guidewire tip (small white arrow) identified by susceptibility artifact. The PRIDE projections along the X and Z directions having peaks (encircled) within the close vicinity of the marker position are clearly seen. (From Patil S, Bieri O, Jhooti P, et al. Automatic slice positioning (ASP) for passive real-time tracking of interventional devices using projection-reconstruction imaging with echo-dephasing (PRIDE). Magnetic resonance in medicine: official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine 2009;62(4): 935-942, with permission.)

Table 1

Characteristics of commonly used MR fluoroscopy sequences.

Contrast weighting	Sequence	Comments
T1/T2	Balanced SSFP (aka, FIESTA, TrueFISP, or balanced FFE)	- Fast - No gadolinium required -At low flip angles, SSFP (or FLAPS) can be used for positive contrast passive tracking of susceptibility markers - Susceptible to B_0 disturbances and blood flow artifacts in setting of slow or turbulent flow
T1+T2*	GRASS, FISP, FFE	- fast, but less SNR efficient than SSFP
T2	TOSSI (RARE, HASTE) TSE	- Flexibility in image contrast - Relatively fast acquisition
T1	FSPGRFLASHT1-FFE	-May be used with IA gadolinium for roadmap guidance

Abbreviations: SSFP – steady-state free precession; FISP - fast imaging with steady-state precession; FFE – fast field echo GRASS - gradient-recalled acquisition in the steady state; RARE – rapid acquisition with relaxation enhancement; TOSSI – T1-insensitive steady-state imaging; TSE – turbo spin echo FSPGR – fast spoiled gradient echo; FLASH – fast low-angle shot; IA – intra-arterial

Table 2

Passive tracking techniques

Passive Tracking Mechanism	Technology	Advantages	Disadvantages
Negative contrast	Markers with susceptibility artifact (material creating signal void)	- Simple - Safe - Lower cost - Negative contrast can be converted to positive (white marker phenomenon)	- Susceptible to motion and flow artifacts when using subtraction - Loss of the anatomical information around the artifact - White marker phenomenon requires prior knowledge of field distortion gradients
Positive contrast	Markers with T1 shortening (e.g., gadolinium)	- Preserves anatomical information around marker	- Positive contrast coating materials unstable - Inability to use lumen of gadolinium filled catheters/need for larger multi-lumen catheters
Nonproton multispectral	Catheters filled with ^{19}F or ^{13}C	High CNR and/or SNR	- Additional hardware needed for non- ^{1}H magnetic resonance - additional software for post processing - ^{1}H-MRI roadmap overlay is obligatory - Decreased spatial and temporal resolution - Requires use of larger multi-lumen catheters
Direct current	Current applied to wire and/or coils create a local field distortion/signal void (negative contrast)	- Simple - Relatively cost-effective - Negative contrast can be converted to positive contrast using subtraction(current on/off) - Coil can also be used for catheter steering and endovascular imaging	- Coil/wires add to outer diameter of catheter - Requires additional hardware - With subtraction technique, time per image doubles - Potential RF and resistive heating of wires/coils - Not commercially available

Table 3

Passive tracking techniques: advantages and disadvantages compared to active tracking

Advantages	Disadvantages
Simple – no additional hardware or software*	Time consuming
Inexpensive	No automatic visualization or plane orientation correction if device tip exits the imaging plane unless using PRIDE sequence
Does not require long wires or connection tocurrent source*	
Safe (no RF or ohmic heating risk)* Can be used at different field strengths	Tracking mechanism cannot be switched on or off*
Easier miniaturization of devices*	Thick slices required to keep device in imaging place may result in superimposition of anatomic structures

* exception: direct current passive tracking