Specialty Area: MR-guided Electro-Physiology Procedures

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Highlights:

- When it comes to following devices, MR is not like X-ray
- Special strategies are needed to follow devices with MR
- Active MR tracking can be fast, robust and performed in three dimensions
- One must address the potential for MR-induced heating when designing devices

Talk Title: Catheters and guidewires for MR-EP

Target Audience: This talk will be of interest to basic scientists and clinicians interested in MR-guided vascular interventions including electrophysiology of the heart.

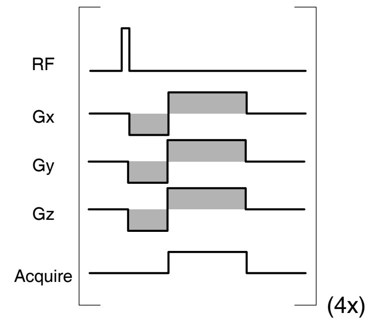
Outcome/Objectives: Attendees of this talk will learn about the challenges of using MR for real-time visualization of devices during MR-guided interventions. Several solutions to real-time tracking, improving tracking robustness and mitigation of rf-induced heating will be discussed.

Purpose/Introduction: Real-time visualization of interventional devices using X-ray fluoroscopy during interventions is reliable, straightforward and fast. While it may be tempting to use similar approaches for MR-guided interventions, the differences between X-ray and MR force very different optimal solutions. With X-ray guidance high-speed projective imaging is performed with a device that is more conspicuous than anatomy. With MR, however, projective imaging is not productive because the complex nature of MR signal (i.e. it has both amplitude and phase) leads to phase cancellation and unpredictable interference patterns along the projection axis. Consequently, most MR imaging is performed with thin slices. Also, imaging frame rates are relatively slow compared to X-ray, particularly if high-resolution is desired. Furthermore, devices such as a catheters or guide wires are typically the least conspicuous feature in an MR image since they have little or no inherent MR signal. The combination of low device conspicuity, slow imaging speeds and thin slice visualization make fluoroscopic MR approaches poorly suited for MR-guided interventions.

Methods

MR Device Tracking: Fortunately, the physics of MR permits alternate approaches to real-time device tracking in MR guided interventions. One strategy that has proven useful is the use of micro-coil detection of MR signals using MR tracking pulse sequences[1,2]. With this approach all the spins in the region of interest are excited, but only those near each micro-coil are detected. Because of this spatial selectivity, only three or four signal acquisitions (i.e. TR periods) are needed to determine the three-dimensional location of the micro-coil. With this non-imaging approach device tracking can be performed at high rates (e.g. 50 frames per second), and over the entire volume of the patient. Since the same principles of physics are used for tracking and imaging, active MR tracking data can be easily registered with conventional MR images. The approach has the added benefit of being relatively quiet.

Figure 1 shows a typical pulse sequence used for MR device tracking of micro-coils. The RF pulse is non-selective (or weakly slice selective) and is applied by the body coil to excite all the spins within the tracking volume of interest. Instead of detecting the signals with the body coil, however, MR signals are detected with small micro-coils incorporated into the device. Since these coils are small, they detect signals only from their immediate neighborhood. Consequently, the signal is a simple peak reflecting the location of the coil along the applied magnetic field gradient (Figure 2). It is important to note that unlike conventional MR imaging, the SNR of the peak is not particularly important. As long as there is sufficient SNR to locate the peak, MR tracking of the device is possible. This obviates the need to tune the micro-coil.



Encoding Schemes

Excitation	X	Y	Z
1	0	0	0
2	1	0	0
3	0	1	0
4	0	0	1

Excitation	X	Y	Z
1	-1	-1	-1
2	1	1	-1
3	1	-1	1
4	-1	1	1

Figure 1. Active MR tracking pulse sequence[2].

When devices are well-constructed and used in well-behaved phantoms, active MR tracking is fast, robust and reliable. Unfortunately, in real-life situations it may not always be possible to match magnetic susceptibility of the device with the patient, or to have an MR signal source in the lumen of the micro-coil, or to have a device which is fully decoupled from other objects in the scanner. Under these conditions active tracking can become challenging.

Fortunately, several approaches have been developed to overcome these complications. The first requirement for micro-coil tracking is to use a robust peak-detection algorithm that can overcome the complex peak shape that can be seen with a solenoid coil. The approach shown in Figure 3 computes the "center of mass" around the highest peak found for a window that is 2.5 times as large as the micro coil. With this method, subtle changes in the peak structure have little influence on the computed peak location, and no *a-priori* knowledge of the coil (other than an estimate of its size) or its orientation in the magnet is needed.

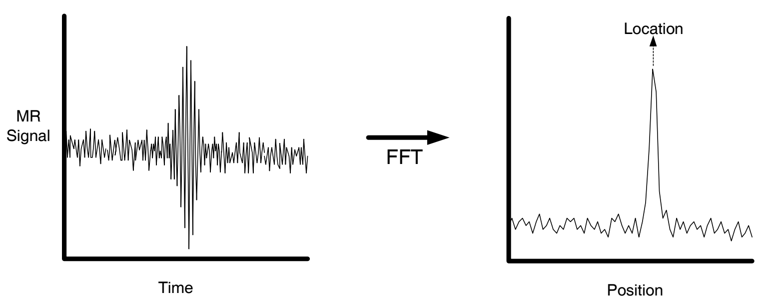


Figure 2. MR signal from an micro-coil before and after Fourier Transformation. Note that the reconstruction yields a simple peak whose location (not amplitude) carries the information needed to compute the location of the micro coil in the device.

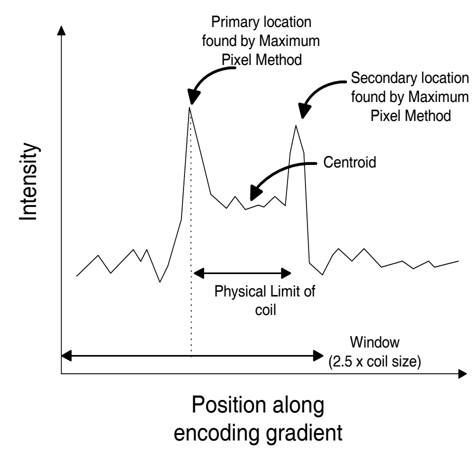


Figure 3. Centroid peak detection algorithm that is insensitive to detail of the peak structure such as the dual peak frequently seen with simple solenoid micro-coils.

Another approach to improve the robustness of micro-coil tracking is to employ a micro-transmitter. Instead of using the body coil to excite all the spins in the body (including those that are of no value for device tracking), the micro-transmit approach uses the micro-coil for both excitation and reception. This offers several advantages including: low rf power (less than 1 Watt) and the elimination of rf-induced heating (see below), and “cleaner” peak profiles.

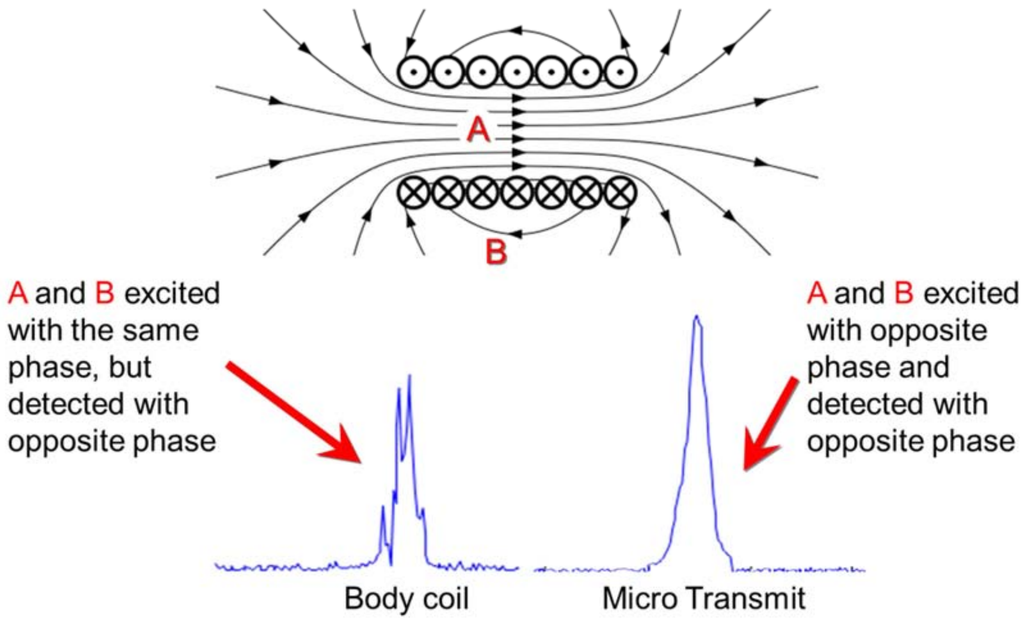


Figure 4. Difference in peak shapes for body coil excitation and micro-transmit excitation. Micro-transmit peaks are “cleaner” because the local field environments (A & B) of a solenoid coil have the same phase for both transmit and excitation.

Often, MR-active catheters and guidewires are constructed with materials that have a magnetic susceptibility that is poorly matched to human tissue. This is particularly true for complex EP catheters[3,4] whose construction can include a variety of different metals and plastics. The consequence of poor magnetic susceptibility matching is that the MR tracking peak can be “smeared out” by local field inhomogeneities and become weaker than the background signal. One strategy to overcome this problem is to employ phase-field dithering[5]. With phase-field dithering, a magnetic field gradient pulse is added to the pulse sequence shown in Figure 1 between excitation and signal detection. The orientation of the dithering gradient is chosen to be perpendicular to the applied frequency encoding gradient. However, there are an infinite number of directions in which this gradient can be applied, and choosing a gradient direction in the absence of any additional knowledge of the local magnetic field environment is just as likely to make the MR signal worse as it is to make it better. With phase-field dithering the entire acquisition is repeated N times with the dithering gradient rotated by 360/N degrees while maintaining its orientation perpendicular to the frequency-encoding gradient. A quality assessment is made for each acquisition and the best peak (e.g. the most intense) is used to compute the location of each micro-coil. While this approach does slow the device tracking frame rate by a factor of N, it does provide robust tracking for devices exhibiting non-ideal behavior such as magnetic susceptibility mis-match and inductive coupling to the body coil. We have found that N=3 offers a reasonable compromise between tracking rate and tracking robustness.

Potential for Device Heating in MR: When a conducting structure such as a guidewire or metallic catheter is used in the dynamic electromagnetic environment of an MR scanner, care must be taken to ensure that dangerous heating of tissue is avoided. Electromagnetic fields can couple to a device to create both common and differential mode currents (Figure 5). In general, differential mode currents are too small to cause concern for micro-coil devices, but common mode currents can have the potential to cause serious harm. Common mode currents become dangerous when they create a standing wave in the electrically conducting structure of the device. Fortunately, several remedies to this problem exist including the use of cable traps (Figure 6), rf filters and high-impedance materials.

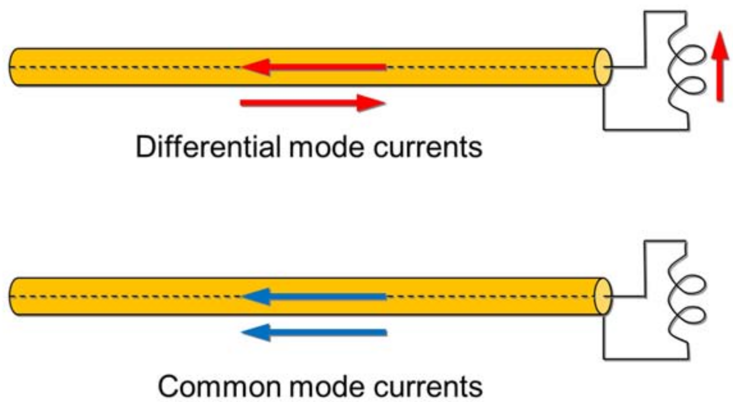


Figure 5. Common mode currents vs. Differential mode currents. In large surface coils measures are taken to block unwanted differential mode currents induced in the coil during rf excitation. For micro-coils, however, these currents are usually too weak to be of concern. Common mode currents, on the other hand can be the source of dangerous heating for MR catheters and guidewires. Fortunately, several strategies can be employed to block these currents.

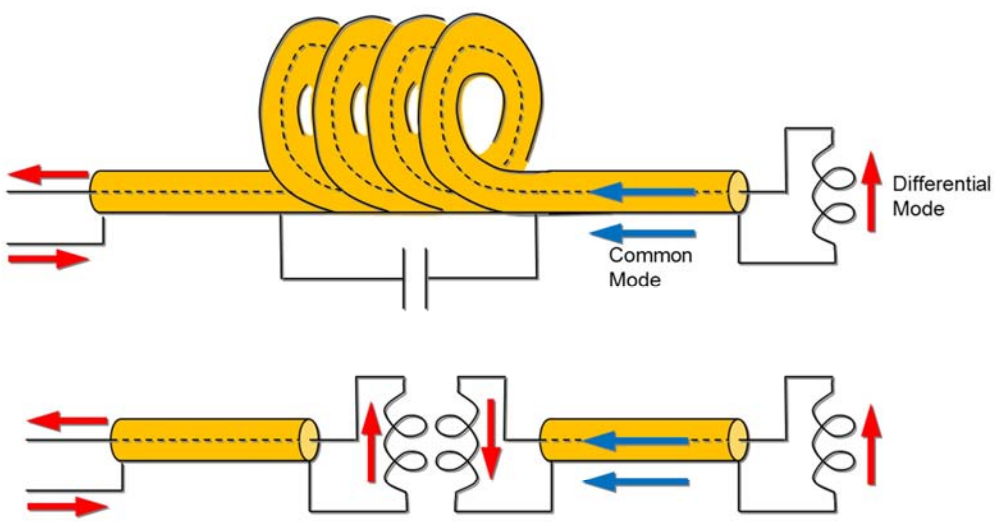


Figure 6. Balun (top) and isolation transformer (bottom) circuits used to block common mode signals which can lead to unwanted heating, while preserving desired differential mode signals (e.g. the MR signal from a micro-coil).

Common mode currents on the conducting structure of a catheter can occur when the structure is exposed to a non-zero E-field. The body coil in modern MR scanners typically has a birdcage design in which the E-field increases proportionally with displacement, “d”, from the center axis of the coil (Figure 7). If the electrical length of the conducting structure is greater than one half of the wavelength, the potential for the creation of a standing wave exists. This situation is difficult to predict since the electrical length of the conducting structure changes as the catheter is inserted and withdrawn in the body. This is because the body’s dielectric constant is different than that of air.

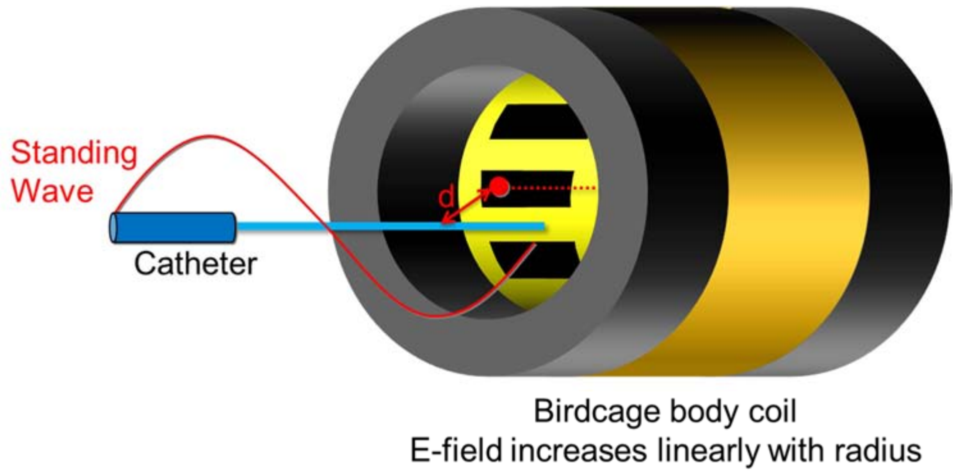


Figure 7. TOP: A schematic catheter placed in the bore of an MR scanner. Note that the catheter is placed off-center by a distance “d”. In this location, the E-field created by the body coil is non-zero. If the electrical length of the catheter is long enough, the conducting nature of the catheter can allow a standing electromagnetic wave to be established. BOTTOM: Collapse of the Tacoma Narrows bridge, November 7, 1940. This is a famous example of a standing mechanical wave that led to a catastrophic failure.

Discussion and Conclusion

The promise of MR-guided interventions is great, but after more than two decades of effort it is still primarily a research tool. If any MR-guided intervention is going to become mainstream, it will need to: A) address an un-met need, B) have wide application, and C) exploit some of the unique strengths of MR. MR-guided Electrophysiology, meets all of these conditions. Although x-ray EP interventions in the heart are becoming common, patients often need to come back for repeat procedures. Clearly, the targeted ablation of cardiac tissue does have proven benefit for patients suffering from cardiac arrhythmias, but many opportunities to optimize its delivery exist. MR-guidance has the potential to greatly improve success rates in EP interventions by providing the interventionist a three-dimensional view of soft tissue and real-time imaging of the therapy (e.g. thermal imaging of ablation). MR guidance has the additional benefit of being radiation-free, which in this context may be of greater concern to the practicing clinician than the patient.

If MR-guided electrophysiology is to become standard practice, a new generation of catheters and sheaths will need to be developed. These devices will first need to be MR-safe. This has major implications in the choice of materials (e.g. non-ferromagnetic metals), and designs that adequately manage rf-induced heating. It is particularly important to match magnetic susceptibility of the device to that of human tissue to permit the acquisition of un-distorted MR signals close to the device.

Another key to success for MR-guided EP will be the ease of following devices in real-time. Unfortunately, treating an MR scanner like an x-ray imaging system is far from optimal. High-speed imaging with MR is relatively slow and low resolution with respect to x-ray. Furthermore, fluoroscopic MR imaging is loud and ambiguities of device location arising from the thin-slice nature of MR and the low conspicuity of devices, make the approach difficult.

Fortunately, the use of active MR tracking in which small receive coils are incorporated into the device overcomes these challenges. Micro-coil tracking can be performed at rates as high as 50 frames per second, in three dimensions, and unambiguously for multiple coils on multiple devices. Strategies to overcome non-ideal behavior such as poor magnetic susceptibility matching and coupling to large coils have been developed and shown to be robust.

While more work is needed to develop MR EP catheters, the initial experience with active MR approaches[3,4] suggests that the benefits may well be worth the investment.

References

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