Magnetic Resonance Imaging in Guidance and Assessment of Cardiovascular Interventions

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1 Introduction

X-ray fluoroscopy is routinely used in patients to guide vascular and cardiac interventions, because of the ability to perform real-time imaging and easy access to patients during interventions (Athanasoulis, 2001; Lakhan et al., 2009; Sousa et al., 2005). X-ray fluoroscopy, however, is limited for defining soft tissue and obtaining functional information. The poor contrast between pathologic and healthy surrounding tissue hinders X-ray fluoroscopy in defining targets (Peters, 2006), which subsequently leads to blind delivery of therapies to the targets (Saeed et al., 2006, 2008a).

On the other hand, MRI uses low energy and no ionizing radiation. It does not require the injection of iodinated contrast, which has been associated with complications, including nephrotoxicity and anaphylaxis. Several studies also showed that exposure to ionizing radiation from X-ray procedures is associated with an increased risk of cancer (Berrington et al., 2004; Frush, 2004; Prasad et al., 2004). A study showed that high dose or repeated administration of gadolinium-based MR contrast media might be a concern, especially in patients with impaired renal function (Sadowski et al., 2007). This problem can be reduced by ensuring a glomerular filtration rate of > 30 ml/min/1.73 m² and contrast agents with high molecular stability (Bongartz et al., 2008).

The recently developed real-time MR sequences offer high temporal/spatial resolution images, safety, accuracy, flexibility and functionality. It also offers rapid recon-
struction and display of 3D images. These features are crucial for minimally invasive vascular and cardiac interventions. Recent improvements in signal processing, tissue characterization and angiographic integration have allowed for MR-guidance in complex interventional procedures (de Silva et al., 2006; Saybasili et al., 2010). Therefore, it has been used in focused ultrasound (Cline et al., 1994, 1995), MRI thermometry (Chung et al., 1999; Cline et al., 1994; Kuroda et al., 2000), functional imaging integrated into MR guided neurosurgical interventions (Yang et al., 2001), local drug delivery (Saeed et al., 2008a; Yang et al., 2006), endoscopy (Hsu et al., 1998), intravascular interventions (Atalar et al., 1998; Bakker et al., 1997, 1998; Ladd et al., 1998; Leung et al., 1995; Smits et al., 1998; van der Weide et al., 1998; Wendt & Wacker, 2000) and intra-operative imaging (Hall et al., 2000; Martin et al., 2000; Samset & Hirschberg, 1999; Schwartz et al., 1999; Yang et al, 2001).

2 MRI Scanners

Open and closed bore MR scanners have been designed for cardiovascular interventions (Hushek et al., 2008). Open scanners were designed to ease patient access/observation and increase comfort for the interventionists. These scanners have low field strength (0.5T), thus offer suboptimal image quality and slow switching speeds that do not meet the need of cardiovascular interventions. Wacker et al. (2005a) found that higher field strength (1.0T) scanners halved the intervention time during stent deployment compared with 0.2T open-bore scanners. The hybrid XMR system consists of an angiographic laboratory adjacent to closed-bore 1.5T MR scanners, wherein an on-track patient table could be moved rapidly between the two imaging modalities (Hushek et al., 2003; Martin et al., 2003). More recently, another XMR hybrid system has been developed that has a side-by-side 1.5T magnet and C-arm X-ray system (Personal communication). The in-suite operation consoles and display monitors are of great help in instant image acquisition and monitoring (Figure 1).

The advantages of hybrid XMR systems are: 1) intermodal movement is minimized because a patient will remain on the same sliding table throughout the intervention and imaging session; 2) unlike single system, the XMR hybrid system permits evaluation of the impact of interventional procedures via MR monitoring; 3) it permits rapid deployment of catheters, and efficient execution of desired interventions without the obligation of using MR compatible devices; 4) it reduces radiation exposure and 5) offers the convenience of a single visit. However, currently XMR systems are available only in few medical centers.

3 Interventional Catheters and Devices

In general endovascular catheters, guide wires and cardiac devices are optimized for their mechanical properties and visibility under projection X-ray imaging. They contain substantial metallic components, such as ferromagnetic material, which is not MR com-
Figure 1: These two XMR suites couple a state-of-the-art MR scanner (background) with a fully functional catheter laboratory (foreground). Hybrid XMR suite equipped with a closed-bore 1.5T MR scanner and C-arm X-ray fluoroscopy. The suite consists of 2 rooms separated by a sliding door. The suite features a floating patient table that can slide a patient quickly and smoothly from one imaging system to the other. (left, Phillips Medical Systems). On the right side, the recently developed hybrid system, where both C-arm X-ray fluoroscopy and 1.5T MR systems are in the same room, thereby making interventional procedures shorter and more efficient (courtesy of Dr. Graham Wright, Sunnybrook, Toronto).

patible. Therefore, their visualization on MRI has been difficult due to susceptibility artifacts derived from the ferromagnetic material, geometry and design (Klemm et al., 2000; Kuehne et al., 2003). Special MR compatible endovascular catheters, guide wires and cardiac devices that are made of nickel-titanium alloy (nitinol), platinum, gold, copper, nonbraided or plastic catheters have been recently developed cause substantially less susceptibility artifacts (Buecker et al., 2004; Kuehne et al., 2002, 2003) and produce less radiofrequency heating in vivo (Nitz et al., 2001). Mekle et al. (2009) used a synthetic MR friendly polymer-based guide-wire for dilatation of an artificial stenosis in phantoms and in the carotid artery, aorta, and iliac arteries of swine. Other investigators manufactured a guide wire based on micropultruded fiber-reinforced material doped with iron particles to improve visibility (Krueger et al., 2008).

Investigators used three approaches for endovascular catheter tracking and navigation, namely passive tracking, active tracking and magnetic catheter steering (Figure 2). The contrast between the catheter and background blood can be improved by injecting MR contrast media, which prevents flow artifacts because steady state is reached earlier (Maes et al., 2005; Martin et al., 2003). Bakker et al. (1997) were the first to use passive tracking approach for steering basilica veins of healthy volunteers. Later, Manke et al and Razavi et al. adapted this passive approach in patients (Manke et al., 2000 Razavi et al., 2003). Contrast media were mounted on non-braided catheters and used as markers for tracking. The advantage of catheter labeling is that it requires no hardware
Figure 2: Selected X-ray photograph (top left), an active MRI catheter (top right) with the coil at the tip (white arrow) embedded in the shaft of the catheter (black arrow) and shoots of MR-guided imaging using passive catheter (bottom left) and active catheter (bottom right). These transendocardial procedures were used to deliver locally different genes or stem cells in infarcted myocardium to enhance angiogenesis and myogenesis.

or instrument modifications and the disadvantage is that the catheter disappears when out of the imaging plane.

Active tracking relies on specially designed micro-coils, electrified wire loop and self-resonant radiofrequency circuits. The coils pick up signal during slice excitation and generate a frequency-encoded recall echo, which can be detected in 3D at a spatial resolution of approximately 1 mm. The micro-coils provide robust tracking of the catheter shaft and tip that allows the user to identify its position and target (Figure 2) (Bock et al., 2004; Saeed et al., 2006; Wacker et al., 2005b). Quick et al. (2002) used antennas for active catheter tracking and imaging of the abdominal aorta, superior mesenteric artery, renal arteries, hepatic artery and celiac trunk.

The safety of active endovascular devices is still a major concern in MR-guided interventions. The conductive nature of the long metallic braids creates a safety hazard
in the MR environment, as the braided shaft can interact with incident RF energy and the electric field transmitted from the RF coil (Park et al., 2007). The heat created by the active coils causes necrosis of the tissue adjacent to the catheter and blood clotting, which may lead to vascular embolization. The methods for mitigating the potential for heating include using unbraided catheters, insulating the conductive structure, limiting the RF power to which it is exposed, or altering its interaction with the RF energy source (Kocaturk et al., 2009). The FDA limits the allowable power deposition via MR imaging to 8 W/kg and temperature change to 2°C.

Magnetic catheter steering is a new approach for tracking endovascular catheters using remote control (Bernhardt et al., 2011; Settecase et al., 2011). This approach relies on a small magnetic moment created by application of an electrical current to copper coils on the catheter tip, which results in alignment of the catheter in the direction of the B0 field. Magnetic catheter steering approach allows for more efficiency in navigating small, tortuous blood vessels, which are currently difficult to catheterize due to buildup of friction at vascular bends. In addition to improved visualization of the endovascular catheter at low power levels, this technology permits deposition of thermal energy for ablation of tissues at higher power levels. This approach is currently under extensive work in our laboratory.

### 3.1 Contrast Media

MR-guided intervention can also benefit from using MR contrast media. MR contrast media represents alternative diagnostic option in patients at risk for adverse reactions to iodinated contrast media. In the early phase of using MR contrast media, the main problem was toxicity because investigators used pure paramagnetic heavy metal ions. Later paramagnetic ions were chelated with DTPA, DOTA or BOPTA to reduce their toxicity at the same time it reduces some of the paramagnetic properties of the free ions. In the late 1980 and early 1990 the first extracellular MR contrast media (Gadopentetate Dimeglumine, Magnevist) was approved for clinical routine. Intravascular contrast media are available for preclinical use only. Extravasation and elimination of intravascular contrast media are very slow compared with extracellular contrast media.

The most common classification of MR contrast media is based on their distribution in the tissue, namely the extracellular (low molecular weight; < 2 kDa), intravascular (high molecular weight; > 50 kDa), and/or intracellular compartment. Unlike extracellular and intracellular, intravascular contrast media remain in the intravascular compartment for a prolonged period due to their size and composition, therefore they provided extended delineation of vascular tree during MR-guided interventions. MR contrast media have been used on MR-guided procedures to improve visualization of devices (Hsu et al., 1998; Saeed et al., 2008a; Yang et al., 2001, 2006) in road mapping blood vessels (Buecker et al., 2004; Maes et al., 2005; Martin et al., 2005; Smits et al., 1998; ) and defining pathologic targets (Hsu et al., 1998; Saeed et al., 2006, 2008a; Samset et al., 1999; Yang et al., 2006).
3.2 Real-Time MRI Sequences

MR-guided interventions became possible because of major advancements in the speed of data acquisition, data transfer, and interactive control and display. Other factors include highly uniform magnetic fields, rapidly changeable magnetic field gradients, multi-channel receivers and computing systems. Real-time MR sequences achieve high speeds by maximizing the switching rates of gradients and RF pulses. The speed of imaging is determined by how quickly spatial encoding can be performed and how fast k-space data can be acquired. Actively shielded, strong, fast-switching gradients and fast electronics have allowed data acquisition intervals to be reduced.

Most modern real-time MR implementations employ balanced steady state free precession techniques because of efficient use of magnetization, high SNR, and short repetition times (Bock et al., 2006; Busse et al., 2001; Duerk et al., 1998; Elgort & Duerk, 2005). The performance of these sequences is currently in the range needed to perform MR guided procedures at >5 frame per second (Bock et al., 2006). The steady state free precession acquisitions have been performed using radial (Peters et al., 2003), and spiral (Spielman et al., 1995) k-space trajectories; ie the readout MR signal is stored in K-space, which is equivalent to a Fourier plane. These acquisition techniques in conjunction with spiral or radial filling of the k-space are considered very reliable for high spatial and temporal resolutions. These imaging sequences also benefit from the use of multiple receiver coil elements (Pruessmann et al., 1999; Rasche et al., 1997). Parallel imaging accelerates acquisition by using the different spatial sensitivities of the coils to correct for under-sampling of image data (Niendorf & Sodickson, 2006). Other sequences that can improve imaging speed while simultaneously balancing image quality include non-Cartesian k-space sampling, temporal data sharing between images, and adjusting the tradeoff between region of interest coverage, temporal and spatial resolution (Elgort & Duerk, 2005). The use of 32 channel receiver arrays that will perform rapid 3D cardiac imaging and parallel transmission techniques to permit more efficient data collection, are also under active investigation (Kyriakos et al., 2006). It should be noted that real-time MR sequences are not free of limitations. For example, the closed configuration of MR scanners >1.5T limit access to the patient and RF pulses induce heating when conductive material is applied in devices; MR imaging is sensitive to magnetic field inhomogeneity, pulsatility/motion of spins and chemical shift.

Pre and post-intervention, the following MR sequences were used: (a) balanced fast field echo cine MR imaging for measuring LV volumes, ejection fraction, cardiac output, stroke volume, LV mass, wall thickness and radial strain (Carlsson et al., 2008; Carlsson et al., 2011; Dicks et al., 2009, 2010; Jacquier et al., 2007; Saeed et al., 2008b, (b) tagged gradient echo planar imaging for measuring circumferential strain and LV rotation (Carlsson et al., 2011; Dicks et al., 2009), (c) phase-contrast velocity-encoded gradient echo planar imaging for measuring longitudinal strain (Bergvall et al., 2006), (d) T2-weighted turbo spin echo sequence for measuring interstitial edema after ablation, (e) T2* multi-echo gradient echo sequence for measuring vascular and myocardial hemorrhage after intervention (Saeed et al., 2010), (f) T1-weighted gradient echo (radiofrequency spoiled) perfusion imaging sequence for measuring myocardial perfusion
changes after delivery of therapy and (g) delayed contrast enhanced T1-weighted gradient echo sequence for assessing tissue viability.

4 Applications of MR-Guidance

4.1 Vascular Interventions

In the last decade MR imaging has been extended from a diagnostic to dynamic modality by tracking intravascular guide-wires and catheters in real-time. It should be noted that only a few investigators have performed vascular stenting in patients under MR guidance (Manke et al., 2001; Paetzel et al., 2005). In 1997 the first human MR-guided study was performed and showed excellent visualization of an endovascular catheter labeled with dysprosium ring markers (Bakker et al., 1997). In this study, investigators did not use guide wires during the movement of the catheter in the cephalic vein of healthy volunteers. Later, MR-guided percutaneous transluminal angioplasty was conducted without complications in 13 patients with iliac stenosis (Manke et al., 2001) and in 15 patients with femoral and popliteal artery stenosis (Paetzel et al., 2005). Furthermore, MR imaging provides detailed information on vascular layers and is able to differentiate between plaque components, such as fibrous, lipid rich and calcified tissue (Choudhury et al., 2004; Kramer et al., 2007).

Stenting and/or angioplasty have been performed using MR-guidance for dilatation of the aorta, pulmonary, coronary, renal iliac and femoral arteries (Boll et al., 2004; Buecker et al., 2000; Choudhury et al., 2004; Dion et al., 2000; Hamer et al., 2006; Kramer et al., 2007; Kuehne et al., 2003; Mahnken et al., 2004; Manke et al., 2001; Omary et al., 2000; Paetzel et al., 2005; Raman et al., 2005; Raval et al., 2005; Saeed et al., 2006; Spuentrup et al., 2002a). Kos et al. (2009) used a polyetheretherketone-based MR imaging-compatible guide-wire for aortic stenting and vena cava filter placement in swine. Several groups have successfully used MR-guidance for placement of vena cava filters (Bartels et al., 2001; Bucker et al., 2001). Pulmonary artery stents have also been accurately implanted across the pulmonary valve (Kuehne et al., 2001, 2002, 2003). Mahnken et al. (2004) used MR-guided procedures for placement of aortic stents grafts and Manke et al. (2001) successfully deployed stents under MR-guidance in iliac arterial stenosis in patients. Post-interventional MR imaging confirmed the location and functionality of the stents.

MR-guidance has also been used for assessment of the pulmonary arterial pressure in pediatric and adult patients with congenital heart disease (Kuehne et al., 2004b; Razavi et al., 2003). A variety of MR-guided interventions have been performed in patients with congenital heart diseases including; placing transjugular, intrahepatic porto-systemic stents, radiofrequency ablation, aortic coarctation, atrio-septal defect and cardiac catheterization. In 2006, Krueger et al. (2006) performed the first MR-guided study using balloon angioplasty for treating aortic coarctation in 5 patients. This was an important step toward MR-guided treatment of congenital diseases.

Kuehne et al. (2004a) demonstrated successful implant of a self-expanding stent
valve in the aorta via percutaneous access under MR-guidance. Transcatheter aortic valve implantation, either retrograde through a transfemoral approach or antegrade through a transapical approach, has become a clinical reality in the treatment of critical aortic stenosis in high-risk patients. MR-guidance plays an important role in transcatheter aortic valve implantation and replacement of insufficient aortic or pulmonic valves (Buecker et al., 2004; Kuehne et al., 2001, 2002). MR imaging enables accurate and reproducible quantifications of regurgitate fraction before and after valve placement. Under MR-guidance, McVeigh et al. (2006) used apical access to guide the placement of a prosthetic aortic valve in beating heart.

### 4.2 Cardiac Interventions

MR imaging is a technique that provides high-resolution 3D images of the heart. Percutaneous closure of atrio-septal defects under MR-guidance has been proven in animals (Rickers et al., 2003), but is hampered by image artifacts produced by the closure devices and the use of fast sequences for cardiac imaging (Shellock & Valencerina, 2005). Atrial septal defect (ASD) is another congenital defect common in children, leading to heart failure and pulmonary hypertension. Percutaneous transcatheter delivery of an ASD occluder has been performed on X-ray fluoroscopy (Omeish & Hijazi, 2001). MR imaging provides reliable diagnosis of ASD (Kersting-Sommerhoff et al., 1990) and MR-guidance was used for delivery (Figure 3) and sizing of ASD closure (Figure 4) (Buecker et al., 2002a, 2002b Schalla et al., 2005). The occluder was made of a nitinol mesh to reduce the distortion of artifacts in the images (Schalla et al., 2003).

Others used a commercial nitinol snare coaxial cathetersystem for delivering septal occluders (Schalla et al., 2005). The advancement of the delivery system through the IVC to the right atrium was monitored under MR-guidance (Figure 3). Other studies used active catheters to approach the left atrium and ventricle from the right atrium and ventricle and measure flow and pressure changes resulted from defects (Razavi et al., 2003; Schalla et al., 2003). Measurements of flow on velocity encoded MR imaging and blood pressure were used to calculate pulmonary resistance. The flow and resistance data obtained on Fick and MR cardiac catheterization methods were in agreement.

A clinical study in 10 patients and 5 volunteers showed that MR-guidance is suited to guide flow directed catheters for measurement of invasive pulmonary artery pressures (Kuehne et al., 2005). Pulmonary vascular flow was noninvasively measured using velocity-encoded cine MR imaging, while pulmonary pressure was measured invasively through a catheter guided into the pulmonary artery under MR-guidance. The results indicate that MR imaging is a promising tool for measurement of pulmonary vascular resistance in patients with different degrees and forms of pulmonary hypertension. MR-guidance has also been used in connecting cardiac chambers and blood vessels in a swine model, where Arepally et al. connected the right and left atrium by puncturing the interatrial septum using an active Brockenbrough-style needle (Arepally et al., 2006). In a clinical study in seven patients, Dick et al. conducted trans-septal puncture and balloon septostomy under MR-guidance (Dick et al., 2005).
**Figure 3:** Percutaneous transcatheter delivery of an ASD occluder device at different phases of expansion in vitro (top) and MR images showing the advancement of the ASD closure into the site of the defect in in vivo (bottom, arrows).

**Figure 4:** MR (left) and X-ray fluoroscopy images (right) were used to measure the diameter of an ASD defect. The images show the close relation between the sizes of the defect on both modalities.
5 MR-Guidance for Delivering Local Therapies

Recently, angiogenic growth factor proteins, gene and stem cell therapies have been delivered, during coronary artery bypass grafting, as an alternative treatment to restore myocytes and blood vessels in end stage patients (Allen et al., 1999; Kleiman et al., 2003; Laham et al., 1999; Ruel et al., 2002; Simons et al., 2000; Stamm et al., 2007; ). Others used catheter-based local delivery approaches (Figure 5).

Figure 5: Contrast enhanced MR image shows the scar in the left ventricle (A, arrows). Real time snapshots of the advancement of the active catheter into the aorta (B), left ventricle (C) and injection of therapy into the target (D). This procedure was used to deliver angiogenic genes and labeled stem cells.

Preclinical and clinical studies have shown that the percutaneous local delivery approach (intramyocardial and intraarterial) is visible under MRI-guidance (Assmus et al., 2006; Freyman et al., 2006; Menasche et al., 2008; Narazaki et al., 2008; Saito et al.,
Animal studies confirmed the success of catheter-based transendocardial delivery of genes in infarcted myocardium. The benefits of catheter-based local delivery of therapies are: 1) targeting only the diseased region, 2) delivering a high local dose, 3) eliminating a high systemic dose and side effects and 4) reducing the chance of angiogenesis in hidden tumor sites especially in elderly patients (Allen et al., 1999; Kleiman et al., 2003; Laham et al., 1999; Ruel et al., 2002; Simons et al., 2000; Simonetti et al., 2002; Stamm et al., 2007).

Preclinical studies have indicated that MR imaging provides quantitative data on myocardial viability, infarct transmurality, microvascular obstruction and hemorrhage. These capabilities have positioned MR imaging as an important approach to pursue for assessing the benefits of locally delivered genes (Saeed et al., 2008a; Yang et al., 2006). Another MR study showed the increase in collateral blood flow of infarcted myocardium after delivering vascular endothelial growth factor, which was confirmed on histology (Figure 6) (Pearlman et al., 2000).

Post et al. (2006) demonstrated an improvement in regional radial strain after intramyocardial injection of adenovirus coding for P39 gene. Furthermore, Liu et al. (2006) found improvement in LV ejection fraction and smaller number of segments with wall motion abnormality after intramyocardial injection of fibroblast growth factor.

Local stem cell transplantation is another therapy for treating ischemic heart disease. The therapeutic effect of stem cells seems to be related to the release of angiogenic factors rather than trans-differentiation of delivered stem cells. Two predominant routes for stem cell delivery to infarcted myocardium are intracoronary infusion and direct intramyocardial injection. Each of these delivery routes attempts to maximize the retention of delivered cells to infarcted myocardium. Early clinical studies indicated that cell transplantation, delivered under MR-guidance, is safe and feasible (Dick et al., 2003, 2005; Hill et al., 2003; Kraitchman et al., 2003). MR imaging has been used not only to track stem cells in the myocardium, but also to non-invasively evaluate ventricular function, perfusion and viability (Ebert et al., 2007). Cell tracking on MR imaging is based on labeling injected cells with FDA approved super paramagnetic iron oxide particles (Budde et al., 2009; Kraitchman et al., 2003). It has been shown that iron labeled cells maintain their viability, proliferation and differentiation (Hill et al., 2003). The cluster of iron labeled cells appear dark on T2* and T2 MR images (Arbab et al., 2005; Budde et al., 2009; Ebert et al., 2007; Kraitchman et al., 2003). Several factors affect the detection of iron labeled cells, which include magnetic field strength, labeling efficiency, type of cells and time of imaging after delivery. Investigators found that the duration of MR detection varies between cells; up to 5 weeks for stem cells (Himes et al. 2004) and up to 16 weeks for skeletal myoblasts (Cahill et al., 2004). Investigators also found hypointense tiny regions far from the site of injection, indicative of migration of stem cells within the infarction several weeks after delivery. MR imaging was used to evaluate changes in LV remodeling following the delivery of cellular therapy (Amado et al., 2005; Arai et al., 2006; Grauss et al., 2008; Hashemi et al., 2008; Moelker et al., 2006; Ziebart et al., 2008). Amado et al. (2006) were able to identify a time-dependent recovery of local contractility associated with the appearance of new tissue resulting from transplantation of allogeneic stem cells in a pig model of infarct.
Figure 6: Histology of scar infarct in control (left) and VEGF gene–treated animals (right) 8 weeks after infarction. Sections A-D were stained with Masson trichrome stain, while E and F were stained with biotinylated isoelectin B4. I = infarction in both groups which is comprised of homogeneous replacement fibrosis with a distinct boundary at the interface between scar and viable myocardium (M). Treated animal (right) contained numerous vessels (arrows), while control animal contained very few vessels. Biotinylated isoelectin B4 localized vessels with brown reaction product, accentuating the neovascularity in VEGF gene–treated infarct animal compared with control animal. LV = left ventricle, calibration bars = 80 μm.
MRI-guided coronary artery stent placement is a challenging interventional procedure because of the small size of the coronary arteries combined with incessant motion during the respiratory and cardiac cycles. These obstacles necessitate higher temporal and spatial resolution for real-time MR imaging techniques when compared with interventional peripheral MR angiography (Spuentrup et al., 2002b). Bock et al. (2008) described in detail the technical prerequisites for MR-guided endovascular interventions and addressed the safety aspects of this technique. The most common complications of coronary PCI are bleeding, hematoma, pseudoaneurysm, and arteriovenous fistulae at the access site.

6 MR-Guidance in Tissue Ablation

Ventricular tachycardia can lead to debilitating symptoms, hospital admissions, implantable cardioverter-defibrillator shocks and death. Radiofrequency catheter ablation of re-entrant circuits within myocardial scar is increasingly used in refractory cases with 50% success rate because of the limited depth of ablation. High-intensity focused ultrasound (HIFU) is an ablative energy source that can precisely penetrate deep into the targeted tissue without affecting surrounding tissues. This technique has been used in conjunction with MR-guidance to provide real-time anatomic and thermal mapping (Ellis et al., 2013). Clinical studies showed atrial scar on contrast enhanced MR imaging that results from RF ablation (McGann et al., 2008; Peters et al., 2007; Reddy et al., 2008). Other studies have demonstrated the association between infarct scar, border-zone and the risk of monomorphic ventricular tachycardia (Bello et al., 2005; Nazarian et al., 2005; Schmidt et al., 2007). Dong et al. (2006) found that 3D MR imaging is helpful for tailoring ablations to the variant pulmonary vein anatomy in 47% of patients with atrial fibrillation. They also noted that 3D images of the atria helped in localizing areas along the tissue ridge separating the left atrium from the pulmonary vein (Dong et al., 2006; Mansour et al., 2006). The ability of real-time MR imaging to visualize the needle tip in the inferior vena cava, atria, fossa ovalis, and surrounding vasculature during transseptal cardiac punctures has also been demonstrated (Arepally et al., 2005; Kenigsberg et al., 2007; Raval et al., 2006). The current interest is to develop a catheter based MRI-guided HIFU technique for ablation of ventricular arrhythmias, but this approach is limited due to the ribs block the energy of the HIFU system.

Recently, we successfully used MRI-guided HIFU in renal ablation and first pass perfusion and contrast enhanced MRI to assess the ablation. Figure 7 shows the perfusion deficits on first pass perfusion imaging of kidneys after ablation, while delayed contrast enhanced MR imaging shows the ablated lesion 10min after contrast administration (Figure 8).

7 Current Limitations and Future Potential

Currently, interventional MRI has several limitations. Installation and operation of MRI equipment is costly. Interventionists require knowledge on MRI and familiarity with the
**Figure 7:** Renal ablation under MR-guided high intensity focused ultrasound (MRg-guided HIFU) using 2 sonication at energy of 4400J per site (left kidney) and 1 sonication per site at energy of 3300J. Perfusion MR images acquired in the first 2 min after bolus injection of 0.2 mmol/kg Gd-DTPA show regional ischemia (arrows). The ischemic lesion is larger on the left than the right kidney.

**Figure 8:** Axial (left) and sagittal (right) contrast enhanced MR images of renal ablation after MR-guided high intensity focused ultrasound (2 sonications showing the lesion in the left kidney (arrows). The deficits were visible for than 45min after contrast injection.
limited scanner bore, while patients must be cooperative during imaging. Transient biological effects have been noted, such as overheating due to alternating magnetic transmissions of the radiofrequency coils. At the present time, most medical and life support devices are MRI incompatible, therefore this technique is limited in acute settings. Thus, care must be taken for patients with aneurysm clips, intra-cranial or intra-ocular metal, shrapnel, cardiac pacemakers or pacemaker wires and cochlear implants. It is important to anticipate the side effects of MRI-guided procedures by using experimental animals before translation into routine clinical practice. Improving safety, spatial and temporal resolution of MRI enhances the feasibility of using interventional MRI in patients. Current, interventional MRI techniques are evolving at a dynamic pace in many centers in the United States and Europe. In the future, the use of contrast media during intervention must be reduced and 3D spatial resolution improved.

7.1 Cost Effectiveness

The potential of interventional MRI is great because as a single modality, it combines 3D anatomic imaging, device localization, hemodynamic/electrophysiologic information, tissue structure and function. The cost of x-ray suite is comparable with MRI suite, but the current costs of most interventional MRI procedures have yet to be reported. Marcy et al evaluated the benefit of performing certain MRI-guided interventional procedures, such as IVC filter placement, percutaneous placement of ureteral stents for obstructive uropathy and percutaneous gastrostomy tube placement. The procedures resolved the symptoms and improved quality of life in at least 80% of patients. The costs of the procedures were relatively small based on the cost of hospitalization (0.85-11.3%) (Marcy et al., 1999). One might expect a negative financial impact, but increasing the utilization of interventional MRI can often result in significant economic benefit. Doubilet et al. (1986) reported that many medical decisions are influenced by factors that cannot be easily quantified or assigned a dollar value. While currently, insurance companies rarely reimburse for these interventional MRI procedures, as they become more routine, they will be more affordable.

8 Conclusion

MR imaging provides 3D datasets, excellent soft-tissue contrast, multi-planar views, dynamic imaging and guidance of interventional vascular and cardiac procedures in a single imaging session. Non-enhanced MR imaging allows noninvasive monitoring of treatment success that is not available on X-ray fluoroscopy. Balloon dilation, stent placement, valvar replacement, atrial septal defect closure, radiofrequency ablation and local gene and cell delivery have been shown to be feasible on MR-guided imaging. It enables a substantially reduced level of invasiveness compared with open-chest surgery, potentially resulting in treatment on an outpatient basis, rapid patient recovery, eliminate radiation exposure and cost savings to the health care system. At present, cardiovascular interventions are addressed by utilizing multimodalities, such as multi-
detector computed tomography, X-ray fluoroscopy and echocardiography. Whether MR is suited to obviate the need for multimodality imaging is currently promising, but unclear. Translation of MR-guided interventions to routine clinical use has been very slow due to limited availability of MR-friendly equipment and funding by National Institute of Health and vendors.

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