Published in final edited form as:

J Magn Reson Imaging. 2005 September ; 22(3): 415–419.

Three-Dimensional Contrast-Enhanced Steady-State Free Precession for Improved Catheter-Directed Coronary Magnetic Resonance Angiography

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Abstract

Purpose—To demonstrate the feasibility of three-dimensional thick-partition, contrast-enhanced, catheter-directed coronary artery magnetic resonance angiography (MRA) and test the hypothesis that three-dimensional imaging improves coronary artery background contrast-to-noise ratio (CNR) compared to two-dimensional imaging.

Materials and Methods—Catheters were advanced into the coronary arteries of swine ($N = 6$) under MR guidance. Three-dimensional coronary MRA was performed after intracoronary injection of a small dose of contrast media using magnetization-prepared steady-state free precession (SSFP) with two thick partitions. For comparison, two magnetization-prepared two-dimensional SSFP scans were also performed, one with no signal averaging and one with two signal averages. All sequences had the same coverage and in-plane spatial resolution.

Results—The coronary artery was successfully catheterized in all (6/6) animals. CNR for three-dimensional imaging was 11.1 ± 1.2 for proximal arterial segments and 4.3 ± 0.4 for distal segments. Without averaging, two-dimensional imaging CNRs for proximal and distal segments were 5.0 ± 0.7 and 1.2 ± 0.2 , respectively. With averaging, two-dimensional imaging CNRs for proximal and distal segments were 9.4 ± 1.5 and 2.9 ± 0.4 , respectively. Three-dimensional imaging showed a statistically significant increase in CNR over all two-dimensional imaging for both proximal and distal segments ($P < 0.05$).

Conclusion—Three-dimensional thick-partition, contrast-enhanced, catheter-directed coronary MRA is feasible and improves CNR over two-dimensional projection imaging.

Keywords

magnetic resonance (MR); coronary vessels; interventional MRI; steady-state free precession (SSFP); MR angiography

MAGNETIC RESONANCE IMAGING (MRI) HAS been proposed as a potential alternative to x-ray fluoroscopy for performing coronary artery interventions (1). MRI-guided procedures have several advantages over x-ray fluoroscopy: the patient and medical staff are not exposed to ionizing radiation, nephrotoxic contrast agent is not used, it is possible to obtain functional as well as anatomical information, and MRI is able to obtain two- and three-dimensional data sets in any arbitrary image orientation. However, MRI-guided interventions are still limited by the achievable spatial resolution, temporal resolution, and vessel signal of the MR image.

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Selective catheter-directed coronary angiography is an essential component of x-ray fluoroscopy-guided coronary procedures. MR angiography (MRA) of the coronary arteries with catheter-directed gadolinium (Gd)-based contrast injections have been shown to be feasible in animals (2). Previous studies have used a two-dimensional thick-slice projection technique with a gradient-recalled echo (GRE) sequence (2,3) or steady-state free precession (SSFP) (4) for data acquisition because imaging time is limited following intracoronary injection of contrast media. Rapid perfusion of contrast agent into the myocardium may obscure the coronary arteries during a prolonged imaging time. However, coronary artery delineation can be poor in two-dimensional projection imaging when the slice thickness is large, even when magnetization preparation or image subtraction is used to reduce background tissue signal because background tissue occupies a much larger volume of a given voxel than the artery of interest.

An alternative approach is to use three-dimensional imaging for catheter-directed coronary MRA, which reduces voxel size and the partial-volume effect. However, three-dimensional imaging requires a much longer imaging time. Contrast agent begins enhancing myocardial signal in as little as 4 seconds following intracoronary injection (4), which can dramatically reduce coronary artery myocardium contrast-to-noise ratio (CNR) if the scan time is too long. This in turn limits the data acquisition window available for coronary artery MRA. A compromise would be to use three-dimensional imaging with a small number of thick partitions to minimize imaging time while maintaining the CNR advantage over two-dimensional imaging.

In this study, we used two partitions for three-dimensional magnetization-prepared SSFP coronary artery MRA after intra-arterial (IA) contrast injection. We tested the hypothesis that this sequence improves coronary artery CNR in swine compared to two-dimensional projection imaging sequence with the same coverage and in-plane spatial resolution.

MATERIALS AND METHODS

Our institutional animal care and use committee approved all experiments, which were conducted in six healthy domestic swine (mean weight = 27.2 kg). All scanning was performed on a 1.5-T Siemens Sonata scanner (Siemens Medical Solutions, Erlangen, Germany) with maximum gradient strength of 40 mT/m. Animal preparation was as follows: On the day of the experiment, the animal was tranquilized with telazol/xylazine and intubated. Following intubation, isoflurane was administered to maintain anesthesia. On-site x-ray fluoroscopy and ultrasound units were used to guide percutaneous insertion of an 8-French vascular sheath into the common femoral artery. A conventional coronary artery catheter was advanced into the coronary ostium under x-ray guidance and baseline coronary x-ray angiography was performed. The coronary catheter was removed prior to subsequent MRI.

In the interventional MRI scanning suite, the coronary artery of the pig was catheterized under MRI guidance through the femoral sheath access site using a 6-French 80-cm-length coronary artery catheter (JL or AL, Cook, Inc., Bloomington, IN, USA) and a coaxially positioned 0.030-inch-diameter guidewire with internal MR coil (Intercept, Surgi-Vision, Inc., North Chelmsford, MA, USA) (5). The catheter was filled with 4% Gd contrast agent (Magnevist, Berlex Laboratories, Wayne, NJ, USA). A Tuohy-Burst-type adapter (Rotating Hemostatic Valve, Guidant Advanced Cardiovascular Systems, Inc., Santa Clara, CA, USA) was applied to the external end of the catheter (outside the body) to prevent leakage of the contrast agent. Under MRI guidance, we tracked the catheter using a real-time inversion recovery (IR)-GRE sequence and the active guidewire using a real-time SSFP sequence described previously (6, 7). Device tracking was performed using a combination of the guidewire internal MR coil and an external phased array body coil. The catheter was then guided into either the left main

coronary artery ($N = 5$) or right coronary artery (RCA) ($N = 1$). After engaging the coronary artery, we verified catheter position by hand-injecting 3-mL aliquots of 4% Gd contrast agent through the catheter, while imaging using IR-GRE with a phased array body coil, closely following a method of MRI-guided coronary artery catheterization described previously (7).

Once the catheter was placed in the coronary arteries, an automated power injector (Spectris, Medrad, Indianola, PA, USA) was used to inject a small bolus of diluted Gd-based contrast agent (8% diluted by volume (8)) through the percutaneously inserted coronary artery catheter for higher-spatial-resolution coronary artery MRA with magnetization-prepared SSFP. Three CE data sets were acquired per experiment using 1) a three-dimensional projection data acquisition (contrast injection duration = 4–6 seconds, injection rate = 1 mL/second), 2) two-dimensional projection acquisition with a reduced imaging time compared to the three-dimensional sequence (contrast injection duration = 2–3 seconds, injection rate = 1 mL/second), and c) two-dimensional acquisition with two signal averages to match the imaging time with the three-dimensional sequence (injection parameters same as 1).

In-plane spatial resolution and through-plane coverage were identical for all MRA sequences. A schematic of the imaging sequence is shown in Fig. 1. All high-spatial-resolution MRA (two- and three-dimensional) was electrocardiographically (ECG) triggered, with data only collected during mid-diastole to reduce blurring due to cardiac motion. Magnetization preparation, which consisted of a 90° saturation pulse followed by a train of four 180° inversion pulses, was played out during the trigger delay period to reduce background signal and increase vessel depiction (9). After the inversion pulse train and immediately prior to data acquisition, a series of five preparation pulses with linearly increasing flip angle were applied to reduce signal oscillations and corresponding image ghosting (10). Data acquisition was centrically encoded in the phase-encoding direction.

A brute-force computer simulation was performed in MatLab (The Mathworks, Natick, MA, USA) to determine the magnetization preparation parameters (TS, TI_1 – TI_4) yielding maximum contrast between CE blood ($T1 \approx 50$ msec) and background tissues, such as fat, myocardium, and blood, at the start of acquisition. In the simulation, the R-R interval was assumed to be 400 msec, corresponding to the smallest R-R interval that would be expected in swine in vivo. The simulated imaging parameters (i.e., TR, flip angle) were chosen based on those used for the in vivo portion of the experiment. Based on simulation results, optimal TS, TI_1 , TI_2 , TI_3 , and TI_4 were determined to be 15, 115, 15, 50, and 72 msec, respectively. Using these parameters, the longitudinal magnetization for all background tissue species was less than 10% at the start of data acquisition.

Typical sequence parameters common to all high-spatial-resolution MRA sequences (both two-dimensional sequence variations and the three-dimensional sequence) were as follows: repetition time (TR)/echo time (TE)/flip angle = 3.6 msec/1.7 msec/ 70° , field of view (FOV) = 113×400 mm², and matrix = 98×512 . To minimize the effects of cardiac motion and heart rate, data were collected in a segmented fashion with typically 33 lines collected per cardiac cycle. For three-dimensional imaging: number of partitions = 2 interpolated to 4, slab thickness = 4 cm. For two-dimensional imaging: slice thickness = 4 cm. We used the magnetization preparation parameters determined by the computer simulation, with the exception of TI_4 , which was adjusted to 95 msec based on empirical results, and this discrepancy is likely due to the fact that certain in vivo tissue types may be more prevalent than others.

To detect an improvement in vessel depiction for three-dimensional imaging over two-dimensional imaging, we calculated relative CNR for each CE data set. Source images were used for all measurements, with a single partition chosen for each three-dimensional data set. Regions of interest (ROIs) were drawn in proximal and distal portions of the coronary artery

for each data set to measure vessel signal. ROIs were also drawn in tissue adjacent to the coronary artery ROI to measure background tissue signal and in air. We calculated relative CNR by subtracting vessel signal minus background signal, dividing by air signal, and multiplying by 1.25 (11). Outcome measures were reported as mean $\times$ standard error (SE) for proximal and distal segments for each technique.

RESULTS

In all cases (6/6), the catheter was successfully guided into the targeted left main or right coronary ostium under MRI guidance for subsequent catheter-directed coronary artery MRA. Following injection of dilute Gd contrast agent, the left main, left anterior descending (LAD), left circumflex (LCX), and/or right coronary artery (RCA) became enhanced (Fig. 2).

Because of the heavy T1 weighting provided by the multiple-IR magnetization preparation, the CE coronary artery signal was high, while surrounding tissue (myocardium, fat) signal intensity was low. Mean CNR $\pm$ SE for the three different imaging techniques examined in this study were calculated and are shown in Table 1. CNR for three-dimensional acquisition was significantly improved over either two-dimensional acquisition scheme ($P < 0.05$). Figure 3 depicts sample comparison images in the same pig to illustrate improved coronary artery delineation using the three-dimensional technique.

DISCUSSION

This study has highlighted the advantages of using thick-partition three-dimensional imaging of the coronary arteries during catheter-directed administration of contrast agent. It has been shown previously that acquisition time is very limited in thick-slice IA coronary MRA because the period between contrast agent injection and myocardial enhancement is very short. We have demonstrated that given a fixed amount of imaging time, acquisition of a three-dimensional data set is feasible and provides superior CNR compared to a two-dimensional sequence with the same coverage and in-plane spatial resolution. The improvement in CNR is even more dramatic for distal segments of the artery (258% increase over two-dimensional imaging without averaging; 48% increase over two-dimensional imaging with averaging), where the reduction in vessel size causes vessel signal to decrease and background tissue can more easily obscure the artery. Because the contrast medium was severely diluted (0.04 mol/liter, compared to 0.5 mol/liter for typical intravenous injection), myocardial enhancement was not observed over the course of the study. This large dilution factor allows for a much larger number of injections (compared to noninvasive MRA with an intravenous injection of Gd) without surpassing the Food and Drug Administration (FDA) limits on administration of Gd contrast agents.

In a thick imaging slice/slab, the major coronary arteries are surrounded by myocardial tissue, epicardial fat, and blood in the cardiac chambers. The dramatic T1 shortening observed following IA injection of contrast agent can be used to depict coronary arteries with high CNR, but only if the background tissue is sufficiently suppressed. A three-dimensional data set with the same in-plane resolution and coverage as a two-dimensional data set will inherently have better coronary artery background tissue CNR because of the partial-volume effect. For an imaging slab with partitions whose thickness is greater than the diameter of a coronary artery, increasing the number of partitions while keeping slab thickness fixed will cause signal in voxels containing only background tissue to decrease more than in voxels containing both background tissue and vessel. As long as partition thickness is greater than coronary artery diameter, increasing the number of partitions should lead to an increase in CNR. However, one consequence of increasing the number of partitions is an increase in imaging time. For IA injections, this typically requires an increase in injection duration (12) to maintain signal-to-

noise ratio (SNR), which can lead to myocardial perfusion by CE blood during acquisition, decreasing vessel-myocardium CNR. This is not as serious a problem for intravenous injections of contrast agent, because the bolus is greatly diluted by the blood pool before reaching the myocardium. Therefore, further improvements in CNR for catheter-directed MRA by increasing the number of partitions may be limited by the increase in imaging time. Fast imaging techniques, such as partial Fourier imaging or parallel imaging, may allow additional partitions to be collected without significant increases in acquisition duration.

Three-dimensional imaging with a small number of partitions typically leads to various image artifacts, such as aliasing and Gibbs ringing. These artifacts were not apparent in this study. We hypothesize that the absence of aliasing artifacts is partly due to the high level of background suppression caused by the magnetization preparation. The magnetization preparation is nonselective, so signal from outside the imaging slab should be quite low, making it unlikely that wrap-around artifacts would decrease image quality. Gibbs ringing is another type of artifact that is often observed when there is a sharp increase/decrease in image intensity over a small number of voxels. This leads to a fluctuation in signal intensity in voxels on either side of the discontinuity. Again, because of the suppressed background, signal intensity oscillations surrounding coronary arteries will not cause major image artifacts.

The safety of catheter-directed MRA has not been validated, and studies aimed at determining the risks of interventional MR devices (such as the loopless guidewire antenna) need to be performed before these techniques can be applied in humans. Although we have demonstrated that three-dimensional MRA improves coronary CNR over two-dimensional MRA, we have not shown that the three-dimensional technique more accurately detects coronary artery stenosis. Future studies that compare the accuracy of catheter-directed MRA and x-ray angiography for detection of coronary artery stenoses are warranted. Additionally, the average human R-R tends to be longer than the average swine R-R. Therefore, the additional number of heartbeats required to acquire multiple partitions using a three-dimensional technique may lead to increased myocardial perfusion (and therefore decreased vessel depiction) in humans compared to swine.

In conclusion, we have shown that three-dimensional imaging schemes can be used for catheter-directed coronary MRA, and that they improve vessel CNR compared to a two-dimensional technique with the same coverage and in-plane spatial resolution. Three-dimensional IA MRA may be useful for identifying coronary artery stenoses in future MRI-guided cardiac interventions.

Acknowledgements

R.A.O. was supported in part by NIH K08 DK60020.

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Figure 1.

SSFP with a saturation pulse-four inversion pulse magnetization preparation. After detection of the R-wave, a 90° saturation pulse is applied, followed by saturation time TS. This is followed by a train of 180° inversion pulses, separated by delay times TI_1 , TI_2 , TI_3 , and TI_4 . Linear flip angle (LFA) preparation pulses are applied at the end of TI_4 . Note that TS and TI_1 – TI_4 are not necessarily equal.

Figure 2.

a: Transverse oblique precontrast localizer of the LAD (arrow). **b:** Thick-slice two-dimensional projection MRA in the same orientation with no signal averages. **c:** Repeat of (b) using two signal averages. **d:** Multi-image projection (MIP) of a three-dimensional catheter-directed CE MRA data set in the same orientation. Marginal branches of the LAD (arrowheads) and distal segments of the artery (dashed arrow) are better depicted in **d**.

Figure 3.

Comparison of coronal two- and three-dimensional images following IA injection of Gd from a catheter preferentially selecting the LAD. **a:** Two-dimensional projection MRA with a short (2-second) scan duration. Some segments of the LAD (dashed arrow) are well depicted, while others are obscured by background signal (solid arrow). **b:** Two-dimensional projection MRA with two signal averages. Increased vessel SNR leads to improved depiction of some segments over (a), but sections of the artery are still obscured. **c:** MIP of three-dimensional data. Due to improved background suppression, CNR over the length of the LAD is improved, and some marginal branches are now visible.

Table 1

Mean Relative CNR Measurements for the Three Different Imaging Schemes for Proximal and Distal Arterial Segments

Acquisition type	Proximal segments (CNR $\pm$ SE)	Distal segments (CNR $\pm$ SE)
3D	11.1 $\pm$ 1.2 ^a	4.3 $\pm$ 0.4 ^a
2D, no averages	5.0 $\pm$ 0.7	1.2 $\pm$ 0.2
2D, two averages	9.4 $\pm$ 1.5	2.9 $\pm$ 0.4

^a3D results that show statistically significant improvement over each of the two 2D imaging schemes ($P < 0.05$).