Temporal Evolution of Susceptibility Artifacts from Coiled Aneurysms on MR Angiography: An In Vivo Canine Study

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BACKGROUND AND PURPOSE: Intracranial aneurysms treated by coiling have a risk for recurrence, requiring surveillance imaging. MRA has emerged as an attractive technique for postcoiling aneurysm imaging. Previous research has evaluated MR imaging artifacts of the coil mass in vitro. Our aim in this study was to evaluate MR imaging artifacts of coiled aneurysms in vivo with time.

MATERIALS AND METHODS: Four sidewall aneurysms were created in each of 4 dogs. Aneurysms were embolized receiving only 1 type of coils. After embolization, the animals were transferred to MR imaging, which included axial 3D TOF MRA (TEs, 3.5, 5, and 6.9 ms), phase-contrast MRA, and coronal CE-MRA. MR imaging studies were repeated at 1, 4, 6, 8, 14, and 28 weeks. We calculated an OEF: $OEF = \frac{V_{\text{artifacts}}}{V_{\text{coil mass}}}$, where the numerator represents the volume of the MR imaging artifacts and the denominator is the true volume of the coil mass measured by 3D RA.

RESULTS: OEFs were largest immediately after embolization and showed a gradual decay until approximately 4 weeks, when there was stabilization of the size of the artifacts. By 4 weeks, there was mild coil compaction (average coil mass volume decrease of 7.8%); however, the OEFs decreased by 25% after 4 weeks ($P < .001$).

CONCLUSIONS: MR imaging susceptibility artifacts change with time, being maximal in the postembolization setting and decaying until 4 weeks. The clinical implications of this study are that baseline MRA for comparison with future imaging should be acquired at a minimum of 1 week after the procedure.

ABBREVIATIONS: ACT = activated clotting time; CCA = common carotid artery; CE = contrast-enhanced; CuSO$_4$ = copper sulphate; 3D RA = 3D rotational angiography; GDC = Guglielmi detachable coil; OEF = overestimation factor; SEM = scanning electron microscopy; TE = echo time; TOF = time-of-flight

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Materials and Methods
For this study, we selected the venous pouch aneurysm in the canine model.20 The model was selected because multiple aneurysms can be created within the same animal, thus permitting a reduction of the number of animals needed. Furthermore, because the aneurysms are located in the neck region, respiratory gating for MR imaging is not necessary. This model has been well-described in the literature as a suitable one for the evaluation of emerging endovascular devices for aneurysmal therapy.21,22

In Vivo Experiments
Anesthesia, Analgesia, and Antibiotic Regimens. All animal experiments were performed in accordance with a protocol approved by our Institutional Animal Care and Use Committee. All procedures were performed with the animal under general anesthesia and by using strict aseptic techniques. Before all surgical or imaging procedures, the animals were preanesthetized by an intramuscular injection of acepromazine (0.06 mg/kg) and glycopyrrolate (0.01 mg/kg). Anesthesia was induced by an IV injection of thiopental (15 mg/kg) or propofol (3 mg/kg) and was maintained with mechanical ventilation using 1%–4% isoflurane. The physiologic status of the animal was assessed by using continuous monitoring of respiration rate, heart rate, oxygen saturation level, end-tidal CO2 level, and temperature. Before surgical procedures, the animals were given buprenorphine (0.02 mg/kg) and a fentanyl patch was applied (50 mg). After surgical procedures, the animals were given buprenorphine (0.02 mg/kg, subcutaneous) and a fentanyl patch was applied (50 mg). The MR imaging technique included coronal T1 unenhanced turbo spinecho, axial 3D-TOF MRA (TEs of 3.5, 5, and 6.9), and phase-contrast MRA (On-line Table). Animals were returned to the MR imaging center at 1, 4, 6, 8, 14, and 28 weeks after coil embolization for repeat imaging studies as previously performed, with the addition of coronal CE-MRA sequences. The in-plane resolution on the MR angiography sequences was approximately 0.4 mm, and section thickness, 0.7 mm. Gadobenate dimeglumine (0.1 mmol/kg, IV) was administered before CE-MRA with bolus tracking; namely, the sequence was commenced on visualization of contrast entering the ascending aorta. Total MR imaging protocol was approximately 30 minutes.

Quantitative Analysis of Susceptibility Artifacts. At each imaging time point, a set of 3D RA images (FOV, 250 × 250 × 250 mm; isotropic voxel size, 0.98 mm) was acquired using an x-ray angiography C-arm system (Allura Xper FD20, Philips Healthcare) to measure the coil mass volume. This method has been shown to reliably determine aneurysm volume23 and is readily applied to measure the volume of the highly radiopaque coil mass. 3D RA measurements were all performed by the same blinded operator, in the same fashion, on dedicated software (XtraVision Release 8.1.1.0002, Philips Healthcare). The measurements were carried out by using a semiautomated method based on local image intensities in the following sequence: 1) The coil mass was isolated by the volume-measuring tool. 2) The threshold was set to the automatic value. 3) The window was set to 100%; whereby creating a binary volume. This was done because the volume-measuring tool uses this setting to determine the volume; changing the contrast has no effect on the volume. 4) The level was visually adjusted to reduce streak artifacts. Because of the relative high intensities caused by the coils, this threshold value was very reproducible with <5% variation between experiments.

To measure artifact volume, MRA sequences were exported to a DICOM file and analyzed in Mimics 13.1 (Materialise, Leuven, Belgium). Once the MR images were imported into Mimics, we chose 2 axial sections: 1 from the proximal set of aneurysms and the other from the distal aneurysms. In each of the sections, a small region of interest measuring 5 mm2 was selected adjacent to (not including) the susceptibility artifacts. Within this small region of interest, the average gray value was calculated. As per the American Society for Testing and Materials 2119–07 guidelines (http://www.astm.org/Standards/F2119.htm) for MR imaging testing of new devices, a value of >30% signal-intensity change was set as artifact. This value was selected as a threshold for outlining the area of susceptibility artifact, and any voxels with a gray value less than this were considered artifact. Manual regions of interest were drawn that encompassed the area of artifact. This manual segmentation was used to limit subsequent thresholding, thereby ensuring that small vessels close to the artifact were not incorporated into the artifact volume measurements. The artifact areas were then reconstructed into 3D volumes, and their volume was measured in cubic millimeters. The OEF was calculated from this volume:24
In Vitro Experiments

To further understand the MR imaging artifacts, we performed a series of in vitro experiments. We created a CuSO₄ gelatin phantom with 4 spheric cavities filled with 0.9% saline; 2 with 3- and 2 with 4-mm radii. To create the phantom, we had a 2-step process: 3-and 4-mm radii spheres were inserted in semiliquid gelatin cups. After the gelatin solidified, spheres were gently extracted from the mold, without collapsing it. This gelatin mold was then filled with 0.9% saline, and the recipient was placed in a −80°C freezer. After the spheres were frozen, they were gently extracted from this mold and embedded in a second CuSO₄ gelatin phantom. After the spheres returned to room temperature, a phantom with 4 saline-filled spheric cavities was obtained.

MR images were obtained at this point to measure sphere sizes. After confirming the size of the cavities, we punctured them and performed a coiling procedure by using identical sizes and types of GDCs or the Target coils with packing densities of 15% and 36% in the larger and smaller cavities, respectively. The same MR imaging protocol described previously was performed, excluding the phase-contrast and contrast-enhanced sequences. The artifact volume was measured as described previously in the canine experiments. The cavities were re-accessed with a 22-ga-long needle for saline evacuation. The coiled cavities were filled with fresh porcine blood. MR imaging and artifact measurements were repeated. 3D RA was performed to record the coil mass size after each MR imaging session.

Statistical Analysis

Results are expressed as mean ± the standard error of the mean. OEFs from longitudinal in vivo studies were analyzed with a repeated-measures ANOVA with a Tukey multiple comparison posttest. OEFs in the phantom models with different environments were compared using a paired Student t test. A value of P < .05 was interpreted as statistically significant.

Results

Twelve canine aneurysms were embolized, and the average packing attenuation was 22.6 ± 7.7%. Four aneurysms were excluded due to complete or partial aneurysmal thrombosis prior to coil embolization. Representative MR images are provided in Fig 1. Figure 2 presents the 3D reconstruction of the susceptibility artifacts (Fig 2A) and 3DRA of the coil mass (Fig 2B). OEFs were the largest immediately after coil embolization and showed a dramatic decrease after 1 week postembolization (P < .01). The OEFs continued to gradually decay until approximately 4 weeks after embolization (P < .001), when there was stabilization of the size of the artifacts (Fig 3). At 4 weeks, we found that there was mild coil compaction (average coil mass volume decrease of 7.8 ± 0.9%). However, the artifacts reduced substantially during the same time period, with the OEFs decreasing by 25 ± 2.8%. At each time point, there was no difference in the OEFs between the different coil families.

Explanted coils from the in vivo aneurysms were examined under SEM to investigate possible galvanic corrosion of the detachment zones, which are constructed from stainless steel (Fig 4). The detachment of the coil is completed by electrolysis of the stainless steel segment. One of the explanted coils had a very small segment of metal exposed (12 μm), and minimal pitting was visualized. However, on a spectral analysis, there was no oxide layer. These results support the hypothesis that galvanic corrosion is not responsible for the decrease in susceptibility artifacts.

Using the phantom, we explored the relationship between MR imaging artifacts and aneurysm volume difference. There
was no difference in OEF between the 3- and 4-mm coiled spheres. The same phantom was used to study the contribution of blood products to the MR imaging artifacts. We had previously measured the artifacts of the coiled cavities with saline. After evacuating the saline and injecting blood, we re-measured the artifacts (Fig 5). From the saline-filled cavity to the blood-filled cavity, there was a 22 ± 2.3% increase in susceptibility artifacts (P < .01).

**Discussion**

In Latin “susceptibilis” is receptiveness. Magnetic susceptibility is a measure of the extent to which a material may be magnetized in relation to a given applied magnetic field. This is a property of the material that can be classified as diamagnetic, paramagnetic, or ferromagnetic on the basis of their susceptibilities. Susceptibility artifacts are generated by an object in the MR imaging with a higher or lower magnetic susceptibility. The presence of any object in the MR imaging scanner will cause a distortion of the main magnetic field. The extent of the distortion is determined by the size and the material properties of the object and the MR sequence used. Paramagnetic objects, such as coils, will cause susceptibility artifacts that may be observed as a shift or loss in signal intensity.

The size and shape of susceptibility artifacts are determined not only by the presence of various materials with different magnetic susceptibilities but also by the shape of these materials. In addition, the shape and size of these artifacts are influenced by the magnitude and direction of the readout gradient. Predicting the shape and size of susceptibility artifacts is a complex task that requires multiparameter simulation.

GDC and Target coils are made of platinum alloy. Previous research has been done in vitro to evaluate MR susceptibility...
Hartman et al. studied GDC artifacts with the coils contained in a plastic vial filled with saline. In their study, it was concluded that the coils produced minimal susceptibility artifacts in all studied sequences (spin-echo T1, fast spin-echo proton attenuation, T2, inversion recovery, TOF) and they described another type of artifact as a 1-mm rim of high signal intensity immediately adjacent to the coil mass. They also evaluated the MR imaging artifacts of 8 patients who were treated with GDC coils and visually compared the images with those from angiography. They reported that there were minimal artifacts, which did not compromise the evaluation of the perianeurysmal area. These high-signal-intensity-rim artifacts have previously been described in other in vitro and in vivo studies. Derdeyn et al. compared MR images with DSA images of 26 patients with intracranial aneurysms treated with coiling. There was no subjective signal-intensity loss beyond the expected margins of the aneurysm; however, no quantitative analysis was performed. Gonner et al. studied an OEF between MRA and DSA images of human coiled aneurysms. However, their OEF was calculated by using the diameter of the artifacts, allowing comparison of 3D MRA with 2D DSA. The overestimation number they described was 2.03 ± 0.88 mm (mean). Although they had different time points at which the MR images were acquired in relation to the coiling procedure, temporal variations of the OEF were not studied. The studies comparing DSA with MRA for follow-up of coiled aneurysm are generally qualitative, comparing parameters such as residual flow within the aneurysm and parent vessel patency. Derdeyn et al. compared MR images with DSA images of 26 patients with intracranial aneurysms treated with coiling. There was no subjective signal-intensity loss beyond the expected margins of the aneurysm; however, no quantitative analysis was performed. Gonner et al. studied an OEF between MRA and DSA images of human coiled aneurysms. However, their OEF was calculated by using the diameter of the artifacts, allowing comparison of 3D MRA with 2D DSA. The overestimation number they described was 2.03 ± 0.88 mm (mean). Although they had different time points at which the MR images were acquired in relation to the coiling procedure, temporal variations of the OEF were not studied. The studies comparing DSA with MRA for follow-up of coiled aneurysm are generally qualitative, comparing parameters such as residual flow within the aneurysm and parent vessel patency. Recently, the use of MR imaging, especially TOF sequences, for posttreatment aneurysm surveillance has been increasingly reported. Generally, a baseline MR imaging is performed before patient discharge, in the same hospital admission. These images are then used as a baseline comparison for future follow-ups. If the same phenomena that we observed happen in humans, we may be overestimating the recanalization rates by reading as recanalization what was previously obscured by artifacts. The data presented herein suggest that baseline postembolization MRA should be performed after waiting at least 1 week following the procedure for comparison with future surveillance studies.

**Conclusions**

In this study, we analyzed the MR imaging susceptibility artifacts generated by bare-platinum-coiled aneurysms at different time points. In this animal model, we observed a 25% increase in the OEF with blood compared to saline. This increase in OEF has important implications for the interpretation of MR imaging studies in patients with coiled aneurysms.
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