Published in final edited form as:

J Magn Reson Imaging. 2017 February ; 45(2): 410–417. doi:10.1002/jmri.25365.

Motion-Insensitive Carotid Intraplaque Hemorrhage Imaging Using 3D Inversion Recovery Preparation Stack of Stars (IR-Prep SOS) Technique

Seong-Eun Kim, PhD^{1,2,*}, John A. Roberts, PhD^{1,2}, Laura B. Eisenmenger, MD², Booth W. Aldred, MD², Osama Jamil, MD², Bradley D. Bolster, PhD⁵, Xiaoming Bi, PhD⁵, Dennis L. Parker, PhD^{1,2}, Gerald S. Treiman, MD^{3,4}, and J. Scott McNally, MD, PhD^{1,2}

¹Utah Center for Advanced Imaging Research, University of Utah, Salt Lake City, Utah, USA

²Department of Radiology, University of Utah, Salt Lake City, Utah, USA

³Department of Surgery, University of Utah, Salt Lake City, Utah, USA

⁴Department of Veterans Affairs, VASLCHCS, Salt Lake City, Utah, USA

⁵Siemens Healthcare, Salt Lake City, Utah, USA

Abstract

Purpose—Carotid artery imaging is important in the clinical management of patients at risk for stroke. Carotid intraplaque hemorrhage (IPH) presents an important diagnostic challenge. 3D magnetization prepared rapid acquisition gradient echo (MPRAGE) has been shown to accurately image carotid IPH; however, this sequence can be limited due to motion- and flow-related artifact. The purpose of this work was to develop and evaluate an improved 3D carotid MPRAGE sequence for IPH detection. We hypothesized that a radial-based k -space trajectory sequence such as “Stack of Stars” (SOS) incorporated with inversion recovery preparation would offer reduced motion sensitivity and more robust flow suppression by oversampling of central k -space.

Materials and Methods—A total of 31 patients with carotid disease (62 carotid arteries) were imaged at 3T magnetic resonance imaging (MRI) with 3D IR-prep Cartesian and SOS sequences. Image quality was determined between SOS and Cartesian MPRAGE in 62 carotid arteries using t -tests and multivariable linear regression. Kappa analysis was used to determine interrater reliability.

Results—In all, 25 among 62 carotid plaques had carotid IPH by consensus from the reviewers on SOS compared to 24 on Cartesian sequence. Image quality was significantly higher with SOS compared to Cartesian (mean 3.74 vs. 3.11, $P < 0.001$). SOS acquisition yielded sharper image features with less motion (19.4% vs. 45.2%, $P < 0.002$) and flow artifact (27.4% vs. 41.9%, $P < 0.089$). There was also excellent interrater reliability with SOS (kappa = 0.89), higher than that of Cartesian (kappa = 0.84).

* Address reprint requests to: S.-E.K., Utah Center for Advanced Imaging Research, 729 Arapleen Dr., Salt Lake City, UT, 84108. sekim@uair.med.utah.edu.

Conclusion—By minimizing flow and motion artifacts and retaining high interrater reliability, the SOS MPRAGE has important advantages over Cartesian MPRAGE in carotid IPH detection.

Carotid artery atherosclerotic plaque is a major cause of stroke, accounting for 10–25% of ischemic strokes in the United States.¹ Studies using magnetic resonance imaging (MRI) suggest that carotid inflammation, intraplaque hemorrhage (IPH), and plaque luminal surface disruption (disruption of fibrous cap), are associated with increased risk of ischemic stroke.^{2–7} Carotid IPH is an important plaque component and can be identified with heavily T_1 -weighted sequences. One T_1 -weighted sequence that has proven useful is the magnetization prepared rapid acquisition gradient echo (MPRAGE), an inversion recovery prepared fast spoiled gradient recalled sequence.^{8,9} Ota et al compared three different T_1 -weighted sequences and concluded that MPRAGE exhibited higher diagnostic capability for the detection and quantification of IPH.¹⁰

The basic premise of the MPRAGE sequence is that tissue with long T_1 relaxation times (blood and muscle) can be differentiated from tissue with short T_1 relaxation times (hemorrhage and fat) using nonselective inversion recovery preparation. Hemorrhage can be further delineated from fat by the addition of fat saturation to the sequence. The inversion time is selected to minimize the signal from flowing blood and other tissues to allow for maximum hemorrhage contrast.

Although promising, MPRAGE accuracy can be limited by patient motion and flow-related artifact. Higher-resolution 3D MPRAGE requires long scan times to compensate for reduced signal-to-noise ratio (SNR), increasing the likelihood of motion artifacts and nondiagnostic studies. A previous study found that over one-third of MPRAGE exams exhibit motion artifacts.¹¹ Elimination of motion artifact will improve IPH detection and better assess IPH location relative to surface disruption. A motion-robust 3D radial and Cartesian hybrid sequence such as 3D “Stack of Stars” (SOS) has been used to acquire volumetric k -space data, where radial sampling is performed in-plane (along k_y and k_x) and Cartesian sampling is used along the slice dimension (k_z).¹² Compared with conventional Cartesian sampling, radial sampling has greater motion robustness because of its varying sampling directions and oversampling of the center of k -space.¹³ The purpose of this study was to compare image quality of the 3D MPRAGE sequence with SOS vs. conventional Cartesian k -space acquisition schemes.

The aim of this work was to determine if the SOS technique would decrease artifacts encountered using the Cartesian MPRAGE sequence, increase image quality, and retain high interrater reliability.

Materials and Methods

Pulse Sequence

The 3D inversion recovery stack of stars (3D IR SOS) sequence was implemented with an inversion recovery preparation and spoiled segmented fast low-angle shot (FLASH) readout capable of Cartesian and SOS trajectories (Fig. 1). A nonselective hyperbolic secant adiabatic inversion RF pulse¹⁴ was applied before each Turbo-FLASH (TFL) readout train

to provide a uniform inversion of spins even in the presence of a nonuniform B1 field and inverts spins within the entire sensitive volume of the transmit coil. Following the IR preparation, all phase-encoding steps along the slice direction (“partitions”) are acquired sequentially before lines at different angular positions (spokes) are acquired. This ensures that the sampling time of the Cartesian trajectory is kept short, which leads to high data consistency within the spoke stack and preserves the motion robustness of radial sampling for the 3D SOS geometry. The Cartesian phase-encoding steps are performed in linear order from $-k_{z\max}$ to $+k_{z\max}$. The same radial or k_y line for all partitions was acquired with a linear k_z -space ordering in the slice direction in the Cartesian trajectory.

In the SOS trajectory, the spokes angles were incremented in a small pseudo-golden angle to acquire eight interleaves in 360° to reduce temporal coherences in k -space. In this case, motion inconsistencies are spread out in k -space and artifacts are visually attenuated. The gradient polarity (readout direction in k -space) of every second spoke was alternated so that adjacent spokes in k -space readout in opposite directions (Fig. 2), which reduces the sensitivity to motion and the visual impact of off-resonance effects.

Signal excitation was performed by a water excitation pulse for optimal vessel outer wall definition and to avoid chemical shift artifacts. The magnetization of blood flowing into the imaging slab is inverted by the IR preparation and should ideally experience the same number of inversion pulses experienced by the stationary tissue.

By using the same imaging parameters including echo time (TE), inversion time (TI), repetition time (TR), and bandwidth (BW), the longitudinal magnetization should be same for both Cartesian and SOS acquisitions. When the magnetization reaches steady state after several repetitions, the longitudinal magnetization can be expressed as:

$$M_z(j) = M_0 \left\{ \left[1 + E_{Td} E_{T0} - 2E_{T0} - \frac{E_{Td} \cos \alpha E_{T0} \cdot (1 - E_1)}{1 - F_{TR}} \right] \cdot (F_{TR})^j + (1 - E_1) \cdot \frac{(1 - (F_{TR})^j)}{1 - F_{TR}} \right\} \quad (1)$$

where $E_{T0} = e^{-\frac{TI_0}{T_1}}$, $E_{Td} = e^{-\frac{TI_d}{T_1}}$, $E_1 = e^{-\frac{TR}{T_1}}$, $F_{TR} = \cos \alpha \cdot e^{-\frac{TR}{T_1}}$ (see Appendix).

Equation (1) was used with scanning parameters matched with our present protocols to simulate and compare the signal behavior of important components including IPH, vessel wall, and blood. For the simulation T_1 values of 500, 1000, 1550 msec, representing hemorrhage, soft tissue (muscle, vessel wall), and blood are compared. In this simulation a linear k -space ordering in the slice direction was used with following parameters; TE/TR = 3.5/10.0 msec, $TI_0/TI_d = 100/50$ msec, flip angle = 11° , and number of slices = 64, and partial Fourier = 6/8. The imaging parameters used in simulation were typical values suggested in other variant MPRAGE techniques.^{9,11,15}

Patient Population

Institutional Review Board (IRB) approval was obtained. Informed consent was obtained from all subjects. In subjects with impaired decisional capacity, legally authorized representative consent and patient assent was obtained. This was a prospective imaging study on patients with known carotid disease. Inclusion criteria included patients with at least 2 mm of carotid plaque on previous clinical imaging. Patients were recruited from neurovascular clinics or inpatient services. Exclusions included patients under 18, prisoners, pregnancy, or those with contraindications to MRI (eg, pacemaker). A total of 31 patients (62 carotid arteries) underwent 3T carotid MRI.

MRI Protocol

Each patient underwent 3T carotid MRI with a custom carotid coil, and IPH was detected by MPRAGE sequence.¹¹ The parameters were: coronal, field of view (FOV) = 160×160 mm², isotropic voxel dimension = 0.73 mm³, TE/TR = 3.5/10.0 msec, 64 slices per slab, partial Fourier factor = 6/8, TI = 350 msec, flip angle = 11°, and two averages. Scan time for SOS trajectory was 5 minutes with 304 spokes per slice. The scan time for Cartesian trajectory was 3 minutes. Other measurement parameters were matched for both SOS and Cartesian acquisitions.

Image Quality and Interrater Reliability Between SOS vs. Cartesian Sequence

The presence or absence of IPH was determined in each of the 62 carotid plaques in the 31 patients undergoing carotid MPRAGE with SOS vs. Cartesian acquisitions at 3T. MPRAGE image quality for both SOS and Cartesian acquisitions was recorded on a scale of 1–5 (1 = poor, 2 = fair, 3 = good, 4 = very good, and 5 = excellent; minimum score = 1, maximum score = 5). Reasons for image degradation were recorded as follows: “motion” = motion degradation, “fat sat” = incomplete local fat saturation resulting in high signal in adjacent fat and sternocleidomastoid, “flow” = high lumen signal due to flow artifact, “signal” = borderline MPRAGE signal due to high noise or low signal. For each reason of image degradation, 1 was subtracted from the maximum score of 5 to give the image quality rating. Interrater reliability was determined between the three observers (B.W.A., L.B.E., J.S.M.).

Statistical Analysis

To determine image quality between SOS vs. Cartesian, mean image quality was first compared between SOS vs. Cartesian using a *t*-test. This was followed by linear regression analysis on the outcome (image quality) and the primary predictor (SOS vs. Cartesian). Finally, a multivariable linear regression analysis was performed with confounders of body mass index (BMI), age, male sex, and disease severity (MPRAGE positive plaque). Kappa analysis was used to calculate interrater reliability. Significant differences in image quality were determined by two-tailed *t*-tests. Differences between percentages of scan limitations (motion, fat sat, flow, and signal) were determined by chi-square tests. All statistical analyses were performed with Stata 13.1 (Statsoft, Tulsa, OK).

Results

Figure 3 shows the simulation results obtained using Eq [1]. The simulated signals from different tissue were compared with inversion times of 250 msec, 350 msec, and 400 msec. Arrows shown in simulated signals at the center of kz where the number of a pulses reaches 32 in Fig. 3a–c. Based on imaging parameters used in the simulation, the inversion time of 350 msec shown in Fig. 3c can provide a reasonable black-blood effect and the best contrast between the hemorrhage and other tissues at the center of k -space. A more important consideration is that this technique must be highly sensitive to hemorrhage, which is characterized by T_1 -shortening in both the intracellular and extracellular methemoglobin state. Figure 3 also shows the simulated magnetization behavior of hemorrhage, muscle, and blood by changing of the flip angle α when the center of k -space has an inversion time of 350 msec (the bottom right-hand side graph, Fig. 3d). Our simulation results showed that for the typical imaging parameters used in MPRAGE acquisition (TE/TR = 3.5/10.0 msec, $TI_0/TI_d = 100/50$ msec, and number of slices = 64 with 6/8 partial Fourier), inversion time of 350 msec and a flip angle of 11° can be used to achieve a reasonable imaging result that is sensitive to detect hemorrhage and to visualize the vessel wall.

A total of 62 carotid arteries were imaged. Representative carotid MPRAGE images obtained from a patient with atherosclerosis and IPH are shown in Fig. 4. The 3D IR-prep-SOS sequence demonstrated excellent blood suppression and clear visualization of IPH relative to wall and lumen. The imaging quality comparison results are summarized in Tables (1–4). In Table 1, image quality (on a scale from 1–5) was significantly higher with SOS compared to Cartesian (mean 3.74 vs. 3.11, $P < 0.001$). The slope of 0.62 in Table 2 indicates the positive correlation between image quality and SOS vs. Cartesian techniques.²¹ After multivariable regression, this high correlation was preserved when controlling for factors commonly associated with decreased image quality, including BMI, age, sex, and disease state (MPRAGE positive plaque) as shown in Table 3. Also, comparing SOS to Cartesian carotid images, motion artifact was decreased (19.4% vs. 45.2%, $P = 0.002$), fat-saturation failure was persistent (51.2% vs. 51.2%, $P = 1.0$), blood flow artifact was decreased (27.4% vs. 41.9%, $P = 0.89$), and signal was less limited (38.7% vs. 64.5%, $P = 0.004$), as shown in Table 4.

Images were rated for IPH with SOS vs. Cartesian MPRAGE as follows (Table 5): B.W.A. rated 23/62 vs. 22/62 as positive, L.B.E. rated 25/62 vs. 24/62 as positive, and J.S.M. rated 26/62 vs. 21/62 as positive. On majority consensus, a total of 25/62 carotid plaques demonstrated IPH on SOS MPRAGE, compared to 24/62 on Cartesian MPRAGE.

Discussion

Our study shows that the 3D IR SOS MPRAGE sequence is a motion-robust and flow-insensitive method for carotid IPH detection. 3D IR SOS images consistently showed decreased motion- and flow-related artifacts. IPH delineation was also much more clearly shown using SOS acquisition compared to Cartesian given the relative preservation of signal. Because the traditional 3D IR Cartesian MPRAGE sequence is the current standard for both carotid MRI research and the clinical standard at multiple institutions,^{9–11} it was

chosen as the control sequence against which to compare the 3D IR-SOS sequence. The quality comparison result in Table 4 shows that motion robustness was significantly higher with SOS compared to Cartesian. The relatively high motion robustness of the SOS sequence mostly derives from the fact that motion causes phase modulations in k -space. During the traditional Cartesian trajectory data acquisition, these motion-related phase variations affect the constant phase differences between the parallel k -space lines and cause severe ghosting artifacts in image space. In the 3D SOS sequence, however, k -space is acquired radially with varying orientation for each line, which means that motion-induced phase effects cannot introduce gaps between adjacent k -space lines that would cause ghosting artifacts. Furthermore, because of the radial geometry, the oversampled k -space, which has an averaging effect due to redundant acquisition of the central information, can add a first-order correction for motion-induced phase errors by adding a constant phase offset to each line to make the phase at the center space the same for all lines.¹⁶ Therefore, motion artifacts appear much more benign as radially oriented streaks and as local blurring of the image, which are less likely to hide pathologic abnormalities and usually do not degrade image quality in a comparable way to ghosting artifacts.¹⁷

There was a positive correlation (regression slope = 0.62) between image quality and SOS vs. Cartesian techniques.²¹ There also was excellent interrater reliability between the three reviewers with SOS (kappa = 0.89), slightly greater than that of Cartesian (kappa = 0.84). The overall quality of 3D IR-SOS images was significantly higher than that of Cartesian images. This is an important step forward for MPRAGE imaging at 3T. A prior study had reported that image quality is better at 1.5T compared to 3T Cartesian MPRAGE, even after controlling for confounders.¹¹ In that prior study, 3T MPRAGE Cartesian images were degraded due to motion, local fat-saturation failure, and flow artifact compared to the 1.5T images. These are known limitations of high-field strength imaging.¹⁸ Since general radial trajectories have higher sampling rates at the center of k -spaces, which results in less time efficiency to cover whole k -space compared to the Cartesian trajectory. To comply with the Nyquist criterion of Cartesian sampling, the radial trajectories often require longer scan time (5 min vs. 3 min). By using the SOS sequence in this study, even the penalty of longer scan time, we have improved upon these sequence limitations. This is important, as carotid MPRAGE is increasingly used in the clinical evaluation of patients with suspected stroke.

A limitation of this study is that motion and flow artifact were not completely eradicated with the SOS modification; however, a further decrease or elimination of these artifacts may be accomplished with cardiac-gating and post-processing in a manner similar to the cineMPRAGE technique.¹⁹ Another limitation of this study was the lack of a true gold standard for IPH, although in previous research we found that Cartesian MPRAGE signal highly correlates with IPH on histology.¹¹ A further limitation of the SOS MPRAGE sequence is that it did not improve local fat saturation failure and this remains a significant contributor to image degradation at 3T. This reduced efficacy of frequency-selective fat-saturation pulses is a known complication of 3T imaging and is secondary to B0 inhomogeneity. These artifacts may be mitigated by improved field shimming and optimal positioning of the patient in the z -direction, since these effects are more pronounced the further the image slice is positioned from isocenter.²⁰

In conclusion, 3T MPRAGE image quality is improved when using SOS compared to Cartesian. With both techniques, there was high interrater reliability, although with SOS it is slightly higher. These results suggest the SOS sequence should be used in preference to Cartesian MPRAGE in both clinical and research applications.

Acknowledgments

Contract grant sponsor: National Institutes of Health (NIH); contract grant number: R01 HL57990; Contract grant sponsor: Clinical Merit Review Grant from the Veterans Administration Health Care System; Contract grant sponsor: Siemens Medical Solutions; Contract grant sponsor: Mark H. Huntsman Endowed Chair; Contract grant sponsor: Ben B. and Iris M. Margolis Foundation; Contract grant sponsor: RSNA Research Scholar Grant; Contract grant sponsor: University of Utah Intramural Seed Grant

Appendix

By using the same imaging parameters including echo time (TE), inversion time (TI), TR, and bandwidth (BW), the longitudinal magnetization should be same for both Cartesian and SOS acquisitions. The longitudinal magnetization following the first IR pulse but before the first excitation RF pulse (α) is given by:

$$M_z(0) = M_0 \left(1 - e^{-\frac{TI_0}{T_1}} \right) - M_1 e^{-\frac{TI_0}{T_1}} \quad (A1)$$

where M_0 is the initial longitudinal magnetization M_1 is the longitudinal magnetization before inversion pulse. TI_0 is the time between the inversion prep and the first excitation pulse. The longitudinal magnetization after the j^{th} excitation RF pulse of the TFL readout is:

$$M_z(j) = M_0 \left(1 - e^{-\frac{TR}{T_1}} \right) + M_z(j-1) e^{-\frac{TR}{T_1}} \cdot \cos \alpha \quad (A2)$$

where TR is time between each α pulse and T_1 is the longitudinal relaxation time.

The longitudinal magnetization M_1 can be written:

$$M_1 = M_z(n) e^{-\frac{TI_d}{T_1}} \cdot \cos \alpha + M_0 \left(1 - e^{-\frac{TI_d}{T_1}} \right) \quad (A3)$$

where $M_z(n)$ is the magnetization after last RF and TI_d is the time between the last (n^{th}) RF and the next IR pulse.

By inserting Eq. [A1] and Eq. [A3] into Eq. [A2], Eq. [2] becomes:

$$M_z(j) = (M_0(1 - E_{T_0}) + M_1 E_{T_0})(F_{TR})^j + M_0(1 - E_1)((F_{TR})^j + (F_{TR})^{j-1} + (F_{TR})^{j-2} + \dots + 1) \quad (A4)$$

where $E_{T_0}=e^{-\frac{T_{I_0}}{T_1}}$, $E_{T_d}=e^{-\frac{T_{I_d}}{T_1}}$, $E_1=e^{-\frac{T_R}{T_1}}$, $F_{TR}=\cos\alpha \cdot e^{\frac{T_R}{T_1}}$ Eq. [A4] can be simplified by using finite geometric series representation:

$$M_z(j)=(M_0(1-E_{T_0})-M_1E_{T_0}) \cdot (F_{TR})^j + M_0(1-E_{TR}) \cdot \frac{(1-(F_{TR})^j)}{1-F_{TR}} \quad (A5)$$

When the magnetization reaches steady state after several repetitions at the end of the train of readout RF pulses, the longitudinal magnetization can be approximated as:

$$M_z(j)=M_z(n)=M_0 \frac{(1-E_1)}{1-F_{TR}} \quad (A6)$$

The longitudinal magnetization at jth pulse can be expressed as:

$$M_z(j)=M_0 \left\{ \left[1+E_{T_d}E_{T_0}-2E_{T_0}-\frac{E_{T_d}E_{T_0}\cos\alpha(1-E_1)}{1-F_{TR}} \right] \cdot (F_{TR})^j + (1-E_1) \cdot \frac{(1-(F_{TR})^j)}{1-F_{TR}} \right\} \quad (A7)$$

References

1. Roger VL, Go AS, Lloyd-Jones DM, Adams RJ, et al. Heart disease and stroke statistics—2011 update: a report from the American Heart Association. *Circulation*. 2011; 123:e18–e209. [PubMed: 21160056]
2. Cai JM, Hatsukami TS, Ferguson MS, Small R, Polissar NL, Yuan C. Classification of human carotid atherosclerotic lesions with in vivo multicontrast magnetic resonance imaging. *Circulation*. 2002; 106:1368–1373. [PubMed: 12221054]
3. Cappendijk VC, Cleutjens KB, Kessels AG, et al. Assessment of human atherosclerotic carotid plaque components with multisequence MR imaging: initial experience. *Radiology*. 2005; 234:487–492. [PubMed: 15671004]
4. Hatsukami L, Ross R, Polissar N, et al. Visualization of fibrous cap thickness and rupture in human atherosclerotic carotid plaque in vivo with high-resolution magnetic resonance imaging. *Circulation*. 2000; 102:959–964.
5. Kerwin WS, O'Brien KD, Ferguson MS, Polissar N, Hatsukami TS, Yuan C. Inflammation in carotid atherosclerotic plaque: a dynamic contrast-enhanced MR imaging study. *Radiology*. 2006; 241:459–468. [PubMed: 16966482]
6. Kwee RM, van Engelshoven JM, Mess WH, et al. Reproducibility of fibrous cap status assessment of carotid artery plaques by contrast-enhanced MRI. *Stroke*. 2009; 40:3017–3021. [PubMed: 19556528]
7. Saam T, Hatsukami TS, Takaya N, et al. The vulnerable, or high-risk, atherosclerotic plaque: noninvasive MR imaging for characterization and assessment. *Radiology*. 2007; 244:64–77. [PubMed: 17581895]
8. Mugler JP 3rd, Brookeman JR. Three-dimensional magnetization-prepared rapid gradient-echo imaging (3D MP RAGE). *Magn Reson Med*. 1990; 15:152–157. [PubMed: 2374495]

9. Zhu DC, Ferguson MS, DeMarco JK. An optimized 3D inversion recovery prepared fast spoiled gradient recalled sequence for carotid plaque hemorrhage imaging at 3.0 T. *Magn Reson Imaging*. 2008; 26:1360–1366. [PubMed: 18583079]
10. Ota H, Yarnykh VL, Ferguson MS, et al. Carotid intraplaque hemorrhage imaging at 3.0-T MR imaging: comparison of the diagnostic performance of three T1-weighted sequences. *Radiology*. 2010; 254:551–563. [PubMed: 20093526]
11. Scott McNally J, Yoon HC, Kim SE, et al. Carotid MRI Detection of Intraplaque Hemorrhage at 3T and 1.5T. *J Neuroimaging*. 2015; 25:390–396. [PubMed: 25040677]
12. Peters DC, Korosec FR, Grist TM, et al. Undersampled projection reconstruction applied to MR angiography. *Magn Reson Med*. 2000; 43:91–101. [PubMed: 10642735]
13. Larson AC, Kellman P, Arai A, et al. Preliminary investigation of respiratory self-gating for free-breathing segmented cine MRI. *Magn Reson Med*. 2005; 53:159–168. [PubMed: 15690515]
14. Silver MS, Joseph RI, Hoult DI. Selective spin inversion in nuclear magnetic resonance and coherent optics through an exact solution of the Bloch-Riccati equation. *Phys Rev A*. 1985; 31:2753–2755.
15. Wang J, Bornert P, Zhao H, Hippe DS, et al. Simultaneous noncontrast angiography and intraplaque hemorrhage (SNAP) imaging for carotid atherosclerotic disease evaluation. *Magn Reson Med*. 2013; 69:337–345. [PubMed: 22442116]
16. Shankaranarayanan A, Wendt M, Lewin JS, Duerk JL. Two-step navigatorless correction algorithm for radial k-space MRI acquisitions. *Magn Reson Med*. 2001; 45:277–288. [PubMed: 11180436]
17. Wech T, Tran-Gia J, Bley TA, Kostler H. Using self-consistency for an iterative trajectory adjustment (SCITA). *Magn Reson Med*. 2015; 73:1151–1157. [PubMed: 24803085]
18. Chavhan GB, Babyn PS, Singh M, Vidarsson L, Shroff M. MR imaging at 3.0 T in children: technical differences, safety issues, and initial experience. *Radiographics*. 2009; 29:1451–1466. [PubMed: 19755605]
19. Mendes J, Parker DL, Kim SE, Treiman GS. Reduced blood flow artifact in intraplaque hemorrhage imaging using CineMPRAGE. *Magn Reson Med*. 2013; 69:1276–1284. [PubMed: 22760825]
20. Dietrich O, Reiser MF, Schoenberg SO. Artifacts in 3-T MRI: physical background and reduction strategies. *Eur J Radiol*. 2008; 65:29–35. [PubMed: 18162353]
21. Stoddard, GJ. *Biostatistics and Epidemiology Using Stata: A Course Manual*. Salt Lake City, UT: University of Utah School of Medicine; 2013. Chapter 2–19 (http://medicine.utah.edu/cccts/sdbc/stoddard_textbook.php)

FIGURE 1.
3D MPRAGE sequence with SOS (left) and Cartesian trajectory (right).

FIGURE 2.
Data sampling pattern of 3D SOS trajectory.

FIGURE 3.

The simulated magnetizations of the tissues with different T_1 relaxation times when inversion time were selected as 250 msec (a), 350 msec (b), and 400 msec (c). The hemorrhage signal was simulated by using different flip angles with TI of 350 msec (d).

FIGURE 4.

Comparison of carotid IPH sequences with SOS acquisition vs. Cartesian acquisition for two different subjects. Flow and motion artifact are eliminated on the SOS acquisition.

TABLE 1

Carotid 3D IR Sequence Image Quality (All Observers)

	Mean	SD	SEM	<i>n</i>	2-tailed <i>P</i>
SOS	3.74	0.83	0.10	62	<0.001
Cartesian	3.11	0.74	0.09	62	

TABLE 2

Linear Regression Model on Image Quality (All Observers)

	Slope	P	95% CI	
SOS vs. Cartesian	0.62	<0.001	0.34	0.90

TABLE 3

Multivariable Linear Regression Model on Image Quality (All Observers)

	Slope	<i>P</i>	95% CI	
SOS vs. Cartesian	0.62	<0.001	0.42	0.82
BMI	−0.01	0.50	−0.05	0.03
Age	0.01	0.61	−0.02	0.03
Male sex	0.49	0.21	−0.28	1.25
MPRAGE positive plaque	−0.02	0.88	−0.30	0.25

TABLE 4

Scan Limitations Between SOS and Cartesian Sequences

	Motion	<i>P</i>	Fat-sat	<i>P</i>	Flow	<i>P</i>	Signal	<i>P</i>	<i>n</i>
SOS	12/62 (19.4%)		32/62 (51.6%)		17/62 (27.4%)		24/62 (38.7%)		62
Cartesian	28/62 (45.2%)	0.002	32/62 (51.6%)	1.0	26/62 (41.9%)	0.089	40/62 (64.5%)	0.004	62

TABLE 5

MPRAGE Positive Plaques by Each Observer for SOS vs. Cartesian

	SOS	Cartesian	<i>n</i>
B.W.A.	23 (37.1%)	22 (35.5%)	62
L.B.E.	25 (40.3%)	24 (38.7%)	62
J.S.M.	26 (41.9%)	21 (33.9%)	62