Published in final edited form as:

Magn Reson Med. 2017 February ; 77(2): 753–758. doi:10.1002/mrm.26111.

Fast Simultaneous Noncontrast Angiography and Intraplaque Hemorrhage (fSNAP) Sequence for Carotid Artery Imaging

Shuo Chen¹, Jia Ning¹, Xihai Zhao¹, Jinnan Wang², Zechen Zhou¹, Chun Yuan^{1,3}, and Huijun Chen^{1,*}

¹Center for Biomedical Imaging Research, Department of Biomedical Engineering, School of Medicine, Tsinghua University, Beijing, China

²Clinical Sites Research Program, Philips Research North America, Briarcliff Manor, New York, USA

³Department of Radiology, University of Washington, Seattle, Washington, USA

Abstract

Purpose—To propose a fast simultaneous noncontrast angiography and intraplaque hemorrhage (fSNAP) sequence for carotid artery imaging.

Methods—The proposed fSNAP sequence uses a low-resolution reference acquisition for phase-sensitive reconstruction to speed up the scan, and an inversion recovery acquisition with arbitrary k-space filling order to generate similar contrast to conventional SNAP. Four healthy volunteers and eight patients were recruited to test the performance of fSNAP in vivo. The lumen area quantification, muscle-blood CNR, IPH-blood CNR, lumen SNR, and standard deviation and intra-plaque hemorrhage (IPH) detection accuracy were compared between fSNAP and SNAP.

Results—By using a low-resolution reference acquisition with 1/4 matrix size of the full-resolution reference scan, the scan time of fSNAP was 37.5% less than that of SNAP. A high agreement of lumen area measurement (ICC =0.97, 95% CI: 0.96–0.99) and IPH detection (Kappa =1) were found between fSNAP and SNAP. Also, no significant difference was found for muscle-blood CNR ($P=0.25$), IPH-blood CNR ($P=0.35$), lumen SNR ($P=0.60$), and standard deviation ($P=0.46$) between the two techniques.

Conclusion—The feasibility of fSNAP was validated. fSNAP can improve the imaging efficiency with similar performance to SNAP on carotid artery imaging.

Keywords

carotid artery; MR angiography; intraplaque hemorrhage; phase-sensitive; SNAP

*Correspondence to: Huijun Chen, Center for Biomedical Imaging Research #109, Tsinghua University, Haidian District, Beijing, China, 100084. chen_hj_cbir@mail.tsinghua.edu.cn.

INTRODUCTION

Carotid atherosclerosis is one of the leading causes of ischemic cerebrovascular events (1). Clinically, the degree of luminal stenosis detected by angiography is the most commonly used index for the detection of high-risk patients (2). In addition to luminal conditions, the morphological and compositional features of atherosclerotic plaque imaged by high-resolution black-blood magnetic resonance imaging (MRI) have widely raised concerns recently (3–5). Several studies have found that intraplaque hemorrhage (IPH) was an important high-risk feature related to ischemic symptoms (6,7). Recently, a simultaneous noncontrast angiography and intraplaque hemorrhage (SNAP) MR imaging sequence was proposed (8), which can provide both features in one scan: stenosis and IPH. The SNAP sequence is a phase-sensitive inversion recovery sequence (9), in which an interleaved full-matrix-size reference scan is used to provide background phase in phase-sensitive reconstruction. Thus, the signal of the tissue with short T1, such as IPH, can be recovered as a hyperintensive positive signal; the negative signal of blood with long T1 can be used to generate noncontrast angiography. However, the acquisition efficiency of SNAP can be potentially improved because of the full-matrix-size reference acquisition that doubles the scan time.

Several techniques have been proposed to accelerate SNAP imaging (10) or phase-sensitive acquisition in general (11–13). One approach used the single center k-space line as reference data to correct the background phase using a linear model (11), but suffered from low accuracy because of the common presence of nonlinear background phase. Recently, a phase-sensitive reconstruction approach without extra acquisition of reference images in inversion recovery sequences was proposed (10). However, such a reconstruction method may cause contrast change and can suffer from phase errors in low SNR regions, which can be commonly found on carotid artery SNAP images.

In this study, we propose a fast SNAP (fSNAP) imaging sequence with full-resolution inversion-prepared acquisition and interleaved low-resolution reference acquisition to speed up the acquisition with similar contrast compared with conventional SNAP. Phantom and in vivo experiments were carried out to demonstrate the feasibility of the proposed fSNAP sequence.

METHODS

Pulse Sequence Design

The design of the fast SNAP sequence is based on the SNAP sequence. A conventional SNAP sequence consists of a slab-selective inversion pulse followed by two turbo field echo (TFE) readout modules (Fig. 1a) with linear k-space filling order with elliptical shutter (Fig. 1b): inversion recovery TFE (IR-TFE) and reference TFE (Ref-TFE), respectively. The Ref-TFE has identical resolution as the IR-TFE, but has a different inversion time (TI), providing the background phase for phase-sensitive reconstruction. Because the variation of background phase should be smooth (14), the proposed fSNAP sequence (Fig. 1c) employs a full-resolution IR-TFE followed by a low-resolution Ref-TFE module. Thus, during one inversion recovery repetition time (IRTR), the IR-TFE module can acquire more data, and

the total scan time will be reduced. In this study, the resolution of Ref-TFE is set to four times lower than that of the IR-TFE. Thus, the data acquired in Ref-TFE compared with IR-TFE is 1:4, resulting in 37.5% scan time saving.

In conventional SNAP, both the IR-TFE and Ref-TFE acquisitions use a linear k-space filling scheme. However, when the linear filling order is used to generate similar contrast (minimizing the sum of square of signal differences in IPH, blood, and vessel wall) to SNAP, the optimized IRTR and TI need to be much shorter (IRTR = 850 ms, TI = 340 ms) with the ratio of IR-TFE to Ref-TFE being 4:1, inversion band coverage = 400 mm, resulting in a much smaller effective imaging coverage (37.4 mm) than SNAP (175 mm) using the established imaging coverage calculation method (8,19). In contrast, the low-high k-space filling scheme with the ability of flexible TI adjustment will reduce the acquisition efficiency. Thus, in this study, we propose the use of the arbitrary k-space filling scheme for fSNAP instead. As shown in Figure 1d, the proposed the arbitrary k-space filling order is implemented by rearranging and cyclic shifting the profile to achieve an adjustable TI for flexible contrast adjustment. To obtain a similar contrast and imaging coverage with original SNAP sequence (IRTR 1980 ms, TI 490 ms, TR 10 ms, FA of IR-TFE/Ref-TFE $11^\circ/5^\circ$), the scan parameters of the fSNAP sequence were set to be similar to SNAP (8) in this study (IRTR 2000 ms, TI 490 ms, TR 10 ms, FA of IR-TFE/Ref-TFE $11^\circ/5^\circ$). Figure 1e shows the signal simulations for the SNAP and fSNAP sequence based on the Bloch equation using MATLAB (the Mathworks, Natick, Massachusetts, USA) with assumed T1 values (blood 1550 ms (15), IPH 500 ms (16), vessel wall 1115 ms (15)) and T2* values (blood 275 ms, IPH 15 ms (17), vessel wall 20 ms (15)). The contrast between the different tissues of fSNAP and SNAP are IPH-blood contrast 0.45 versus 0.44, and wall-blood contrast 0.22 versus 0.2. Moreover, because the IRTR and TI have not been changed much, the effective imaging coverage of fSNAP (175 mm) is the same as the conventional SNAP.

Phantom Experiment

To demonstrate the feasibility of phase-sensitive reconstruction using a low-resolution reference scan, a phantom experiment was performed on a 3T whole body scanner (Achieva TX, Philips Healthcare, Best, the Netherlands) with an eight-channel brain coil. A water phantom with a measured T1 value of 3360 ms was used. A slab-selective inversion pulse with 50-mm longitudinal coverage was used to approximate the contrast between blood and other tissues in SNAP sequence (Figs. 2a and 2e). The imaging parameters for the phase-sensitive inversion recovery with full-resolution reference scan were TI 10.3 ms, TR 10 ms, IRTR 3028 ms, FA $11^\circ/5^\circ$, FOV $250 \times 120 \times 45 \text{ mm}^3$, voxel size $0.8 \times 0.8 \times 0.8 \text{ mm}^3$, TFE factor for IR acquisition and reference acquisition 150/150, coronally oriented, scan time of 168 s. The imaging parameters for the phase-sensitive inversion recovery with low-resolution reference scan were the same as the full-resolution reference, except for TFE factor for IR acquisition and reference acquisition of 240/60, resulting scan time of 105 s. The down-sampled low-reference k-space data were then zero-padded to the full matrix size, followed by conventional reconstruction online.

In Vivo Experiment

Four healthy volunteers (four males, mean age: 24.5 years) and eight patients with atherosclerotic carotid plaque diagnosed by duplex ultrasound (three males, mean age: 71.6 years) were recruited for this study. Written informed consent was obtained from all of the participants before the scan. All in vivo images were acquired on a 3T scanner (Achieva TX, Philips Healthcare, Best, the Netherlands) with a custom-designed 36-channel cardiovascular coil (18). Both fSNAP and SNAP images were scanned in the oblique coronal orientation centered at the carotid artery bifurcation of the index side (as identified by ultrasound) for the patient and left side for the healthy volunteer. The fSNAP and SNAP sequence shared the same TI 490 ms, TR 10 ms, and FA 11°/5°. The only difference was that fSNAP had the TFE factor of 160/40 (IRTR 2000 ms) for IR-TFE/Ref-TFE, whereas SNAP had the TFE factor of 98/98 (IRTR 1980 ms). Other scan parameters for both sequences were FOV of $250 \times 160 \times 32 \text{ mm}^3$, voxel size $0.8 \times 0.8 \times 0.8 \text{ mm}^3$, fat sat, NSA 2. All images were reconstructed online in a Phi-lips scanner. The scan time of fSNAP was 255 s, compared with 408 s in conventional SNAP; a total reduction of 153 s (37.5%) was achieved.

Data Analysis

For the phantom experiment, the corrected phase images were compared between the phase-sensitive reconstructions with high-resolution and low-resolution reference. In the in vivo experiment, the performance of lumen area assessment and IPH detection were compared between the fSNAP and SNAP sequence. For comparison of the lumen area quantification, all acquired coronal images were reformatted into axial slices. Next, the reformatted axial fSNAP slices were matched to the axial SNAP slices using carotid bifurcation as a reference before image review. After that, five slices (two on common carotid artery, one at bifurcation, and two on internal carotid artery) centered at the carotid bifurcation with a 4-mm gap were chosen from bilateral carotid arteries for both SNAP and fSNAP images of each case (in total, 10 slices were chosen for each case; 120 slices from 12 subjects). The bright blood MRA images were generated (8) for both techniques. The lumen of the carotid artery and a uniform region (adjacent sternocleidomastoid muscle in this case) near the artery on those slices were outlined by a reviewer using ImageJ (version 1.34s, NIH, Bethesda, Maryland, USA). The contrast-to-noise ratio (CNR) between the adjacent sternocleidomastoid muscle and blood was calculated with the standard deviation of muscle region as noise (19). Then, the lumen areas derived from the lumen contours and the CNR of the SNAP and fSNAP images were compared using intraclass correlation coefficient (ICC) and two-tailed paired t-tests, respectively. The signal-to-noise ratio (SNR) (20) and standard deviation of lumen were also reported and compared using two-tailed paired t-tests to evaluate the MRA quality. To compare the performance of fSNAP in IPH detection, another review was carried out to identify the hyper-intense signal as IPH on all of the images of 16 arteries from eight patients for both fSNAP and SNAP images using a Philips DICOM viewer (R3.0 SP03). Per-artery based, the Cohen's Kappa was used to determine the agreement of IPH detection on SNAP and fSNAP images. Then the number of axial slices with IPH detected by SNAP and fSNAP were reported for each IPH and compared using Pearson correlation. In addition, the contrast (CNR) between IPH and blood of SNAP and fSNAP were reported. All of the reviews were done blinded to the patient information and

acquisition methods. All of the statistical analyses were performed using Medcalc (version 11.4.2.0, MedCalc Software, Mariakerke, Belgium).

RESULTS

Phantom Experiment

As shown in Fig. 2, the magnitude, phase, and phase-corrected (CR) images were found to be very similar between the phase-sensitive reconstructions with the full-resolution and low-resolution reference scan. The phase of noninverted and inverted regions was consistent between the two techniques. As shown in Fig. 2h, in the curve of a center line of the corrected phase images, both the phase and boundaries of the inversion band had good agreement.

In Vivo Experiment

A typical artery with luminal stenosis and IPH is shown in Fig. 3. Both SNAP and fSNAP can generate MRA of carotid artery with similar quality (Figs. 3a and 3d). Moreover, the IPH can be identified clearly on the SNAP and fSNAP sequence (Figs. 3b and 3c). In the selected 120 locations for lumen size and muscle-blood CNR evaluation, a high agreement was found between the fSNAP and SNAP in lumen-size measurements (ICC =0.97, 95% CI: 0.96–0.99) (Fig. 4a). There was no significant difference of muscle-blood CNR between fSNAP and SNAP ($P=0.25$) (Fig. 4b). Compared with SNAP, the fSNAP demonstrated no statistical difference in lumen SNR (fSNAP: 25.45 ± 6.80 ; SNAP: 25.82 ± 7.68 , $P=0.60$) and lumen signal standard deviation (fSNAP: 167.28 ± 36.25 with mean lumen signal -707.90 ± 215.12 ; SNAP: 161.20 ± 46.44 with mean lumen signal -692.17 ± 211.41 , $P=0.46$). In the 16 arteries from eight patients, both fSNAP and SNAP detected five identical arteries with IPH (Kappa =1, Standard error: 0, 95% CI: 1.0–1.0) (Fig. 4d). Of the five IPHs detected in eight patients, fSNAP and SNAP had a similar number of axial slices that could be detected with IPH (fSNAP: 3, 5, 10, 10, 17; SNAP: 3, 5, 11, 10, 14 slices for each IPH; $r=0.97$; $P=0.006$). The mean IPH-blood CNR of SNAP and fSNAP were 50.77 and 47.72, respectively; no significant difference was found ($P=0.35$).

DISCUSSION

In this study, the feasibility of the fast simultaneous non-contrast angiography and intraplaque hemorrhage sequence was demonstrated. The results of the phantom and in vivo experiments has shown that using a low-resolution reference acquisition for phase-sensitive reconstruction can reduce the scan time by 37.5% without significantly sacrificing the image quality and contrast. More significantly, the proposed fSNAP technique has shown good agreement with the original SNAP sequence in the measurement of lumen area and detection of IPH in a population including healthy volunteers and patients; this indicates its potential to be used as a noncontrast MRA technique and an IPH detection technique in clinical practice. The key for fSNAP to achieve a similar performance to SNAP is the arbitrary k-space filling scheme used in the IR-TFE module of fSNAP. Different from the linear or low-high k-space filling scheme, the arbitrary k-space filling scheme allows flexible TI and IRTR selection to adjust the contrast of fSNAP, without sacrificing imaging efficiency or coverage.

The phase-sensitive reconstruction for fSNAP used in this study is just a simple zero padding of low-resolution reference acquisition with conventional reconstruction online. The simple reconstruction allows fSNAP to be easily transferred to clinical use. Because there is no significant CNR decrease, the fSNAP technique can be further sped up by using other fast imaging methods, such as the parallel imaging SENSE (21) and GRAPPA (22). Moreover, the results of this study suggest that the scheme of fSNAP (the low-resolution Ref-TFE module combined with arbitrary k-space filling IR-TFE module) has great potential to be used in phase-sensitive inversion recovery sequence (PSIR) for brain (23) and cardiac imaging (24). Especially, the ability of flexible TI and IRTR in fSNAP technique could be a great advantage in other applications because of its convenience in contrast adjustment.

One limitation of this study is the lack of validation with histology, because of its unavailability. As a compromise, the histological validated SNAP sequence (8) was compared in this study. Another possible limitation of fSNAP is the Gibbs ringing artifact resulting from the signal truncation at the k-space border in low-resolution reference acquisition (25,26), especially when the reference resolution is too low. However, as shown in the phantom and in vivo experiments, no obvious artifacts are seen with the 1/4 resolution Ref-TFE used in this study. Moreover, although the lumen SNR and signal standard deviation were similar for fSNAP and SNAP, the MRA generated by both techniques showed inhomogeneous signal (Figs. 3a and 3d). The main reason is the dephasing of flowing blood, which can potentially be improved by the flow compensation technique (27). Besides, the present of severe stenosis and calcification may affect the performance of fSNAP, which cannot be investigated in this study because of its unavailability. Thus, the performance of fSNAP should be further investigated in patients with more severe stenosis and calcifications in future studies with a larger population.

In conclusion, an fSNAP sequence was proposed and validated on carotid artery in this study. By using the arbitrary k-space filling order and low-resolution reference scan, fSNAP can improve the imaging efficiency with similar performance in lumen size assessment and IPH detection on carotid artery imaging.

References

1. Takaya N, Yuan C, Chu B, et al. Association between carotid plaque characteristics and subsequent ischemic cerebrovascular events: a prospective assessment with MRI—initial results. *Stroke*. 2006; 37:818–823. [PubMed: 16469957]
2. Moore WS, Barnett HJ, Beebe HG, et al. Guidelines for carotid endarterectomy: a multidisciplinary consensus statement from the ad hoc committee, American Heart Association. *Stroke*. 1995; 26:188–201. [PubMed: 7839390]
3. Yuan C, Mitsumori LM, Ferguson MS, Polissar NL, Echelard D, Ortiz G, Small R, Davies JW, Kerwin WS, Hatsukami TS. In vivo accuracy of multispectral magnetic resonance imaging for identifying lipid-rich necrotic cores and intraplaque hemorrhage in advanced human carotid plaques. *Circulation*. 2001; 104:2051–2056. [PubMed: 11673345]
4. 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]

5. Saam T, Hatsukami TS, Takaya N, Chu B, Underhill H, Kerwin WS, Cai J, Ferguson MS, Yuan C. The vulnerable, or high-risk, atherosclerotic plaque: noninvasive MR imaging for characterization and assessment. *Radiology*. 2007; 244:64–77. [PubMed: 17581895]
6. Singh N, Moody AR, Gladstone DJ, Leung G, Ravikumar R, Zhan J, Maggisano R. Moderate carotid artery stenosis: MR imaging-depicted intraplaque hemorrhage predicts risk of cerebrovascular ischemic events in asymptomatic men. *Radiology*. 2009; 252:502–508. [PubMed: 19508983]
7. Takaya N, Yuan C, Chu B, et al. Presence of intraplaque hemorrhage stimulates progression of carotid atherosclerotic plaques: a high-resolution magnetic resonance imaging study. *Circulation*. 2005; 111:2768–2775. [PubMed: 15911695]
8. Wang J, Bornert P, Zhao H, et al. Simultaneous noncontrast angiography and intraplaque hemorrhage (SNAP) imaging for carotid atherosclerotic disease evaluation. *Magn Reson Med*. 2013; 69:337–345. [PubMed: 22442116]
9. Kellman P, Arai AE, McVeigh ER, Aletras AH. Phase-sensitive inversion recovery for detecting myocardial infarction using gadolinium-delayed hyperenhancement. *Magn Reson Med*. 2002; 47:372–383. [PubMed: 11810682]
10. Wang J, Chen H, Maki JH, Zhao X, Wilson GJ, Yuan C, Bornert P. Referenceless acquisition of phase-sensitive inversion-recovery with decisive reconstruction (RAPID) imaging. *Magn Reson Med*. 2014; 72:806–815. [PubMed: 24407614]
11. Park HW, Cho MH, Cho ZH. Real-value representation in inversion recovery NMR imaging by use of a phase-correction method. *Magn Reson Med*. 1986; 3:15–23. [PubMed: 3959881]
12. Borrello JA, Chenevert TL, Aisen AM. Regional phase correction of inversion-recovery MR images. *Magn Reson Med*. 1990; 14:56–67. [PubMed: 2352472]
13. Xiang QS. Inversion recovery image reconstruction with multiseed region-growing spin reversal. *J Magn Reson Imaging*. 1996; 6:775–782. [PubMed: 8890016]
14. Noll DC, Nishimura DG, Macovski A. Homodyne detection in magnetic resonance imaging. *IEEE Trans Med Imaging*. 1991; 10:154–163. [PubMed: 18222812]
15. Noeske R, Seifert F, Rhein KH, Rinneberg H. Human cardiac imaging at 3 T using phased array coils. *Magn Reson Med*. 2000; 44:978–982. [PubMed: 11108638]
16. 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 Imag*. 2008; 26:1360–1366.
17. Zhu DC, Vu AT, Ota H, DeMarco JK. An optimized 3D spoiled gradient recalled echo pulse sequence for hemorrhage assessment using inversion recovery and multiple echoes (3D SHINE) for carotid plaque imaging. *Magn Reson Med*. 2010; 64:1341–1351. [PubMed: 20574968]
18. Wang, X., Li, R., Hayes, C., Balu, N., Zhao, X., Yuan, C. A new designed 36-channel neurovascular coil at 3T. *Proceedings of the 20th Annual Meeting of ISMRM*; Melbourne, Australia. 2012. p. 2787
19. Wang J, Ferguson MS, Balu N, Yuan C, Hatsukami TS, Bornert P. Improved carotid intraplaque hemorrhage imaging using a slab-selective phase-sensitive inversion-recovery (SPI) sequence. *Magn Reson Med*. 2010; 64:1332–1340. [PubMed: 20597120]
20. Zhou Z, Li R, Zhao X, He L, Wang X, Wang J, Balu N, Yuan C. Evaluation of 3D multi-contrast joint intra- and extracranial vessel wall cardiovascular magnetic resonance. *J Cardiovasc Magn Reson*. 2015; 17:41. [PubMed: 26013973]
21. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med*. 1999; 42:952–962. [PubMed: 10542355]
22. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med*. 2002; 47:1202–1210. [PubMed: 12111967]
23. Hou P, Hasan KM, Sitton CW, Wolinsky JS, Narayana PA. Phase-sensitive T1 inversion recovery imaging: a time-efficient interleaved technique for improved tissue contrast in neuroimaging. *AJNR Am J Neuroradiol*. 2005; 26:1432–1438. [PubMed: 15956512]

24. Kellman P, Arai AE, McVeigh ER, Aletras AH. Phase-sensitive inversion recovery for detecting myocardial infarction using gadolinium-delayed hyperenhancement. *Magn Reson Med*. 2002; 47:372–383. [PubMed: 11810682]
25. Gottlieb D, Shu CW. On the Gibbs phenomenon and its resolution. *Siam Rev*. 1997; 39(4):644–668.
26. Archibald R, Gelb A. A method to reduce the Gibbs ringing artifact in MRI scans while keeping tissue boundary integrity. *IEEE Trans Med Imaging*. 2002; 21:305–319. [PubMed: 12022619]
27. Haacke EM, Lenz GW. Improving MR image quality in the presence of motion by using rephasing gradients. *AJR Am J Roentgenol*. 1987; 148:1251–1258. [PubMed: 3495155]

FIG. 1.

Sequence diagram and k-space filling scheme diagram of the SNAP sequence (**a, b**) and fSNAP sequence (**c, d**). (**a**) SNAP sequence consists of an inversion pulse and two TFE shots (IR-RFE and Ref-TFE) with identical phase-encoding gradients (linear order) but different flip angles for phase sensitive reconstruction. (**b**) k-Space filling order of one shot (red dots) in both IR-TFE and Ref-TFE is also shown (blue line). (**c**) fSNAP sequence consists of two TFE shots with different phase-encoding scheme following the IR pulse. The Ref-TFE acquisition has a low-resolution; the IR-TFE acquisition has an arbitrary order for flexible inversion time (TI) adjustment. (**d**) Arbitrary k-space filling order of one shot (red dots) in both IR-TFE and Ref-TFE is also shown (green line). (**e**) Signal evolution curves of SNAP (IRTR 1980 ms, TI 490 ms, TR 10 ms, FA 11°/5°) and fSNAP (the reduction factor for Ref-TFE is 4, IRTR 2000 ms, TI 490 ms, TR 10 ms, FA 11°/5°) based on the Bloch equation. The contrast among different tissues of fSNAP and SNAP are similar: IPH-blood contrast 0.45 versus 0.44, wall-blood contrast 0.22 versus 0.2.

FIG. 2.

Phantom images obtained by the phase-sensitive reconstruction with full-resolution reference ((a) magnitude image; (b) corrected phase; (c) phase-corrected (CR) image) and low-resolution reference ((e) magnitude image; (f) corrected phase; (g) phase-corrected (CR) image). (d) Difference image between the corrected phase image of the two techniques. (h) Curves are phase of lines in the center of the corrected phase images. Phase-sensitive reconstructed images of the two methods have good agreement.

FIG. 3.

fSNAP and SNAP images of one patient. Comparable MRA and vessel wall images were acquired by both SNAP sequences (**a**) (**b**) and the proposed fSNAP sequence (**d**) (**c**). Both lumen stenosis and IPH can be clearly delineated by both techniques. Yellow arrow: carotid artery stenosis; red arrow head: intraplaque hemorrhage.

FIG. 4.

Statistical comparison of lumen area (**a**) and tissue-blood CNR (**b**) between fSNAP and SNAP. A high agreement (ICC =0.97) was found between the two techniques on lumen area measurement (**a**); paired t test showed comparable tissue-blood CNR between fSNAP and SNAP. The mean and standard deviation are shown as the column and bar (**b**).