Published in final edited form as:

Magn Reson Imaging. 2009 January ; 27(1): 13–22. doi:10.1016/j.mri.2008.05.009.

HASTE Sequence with Parallel Acquisition and T2 Decay Compensation: Application to Carotid Artery Imaging

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Abstract

T2 weighted carotid artery images acquired using the TSE sequence frequently suffers from motion artifacts due to respiration and blood pulsation. The possibility of using HASTE sequence to achieve motion-free carotid images was investigated. The HASTE sequence suffers from severe blurring artifacts due to signal loss in later echoes due to T2 decay. Combining HASTE with parallel acquisition (PHASTE) decreases the number of echoes acquired and thus effectively reduces the blurring artifact caused by T2 relaxation. Further improvement in image sharpness can be achieved by performing T2 decay compensation before reconstructing the PHASTE data. Preliminary results have shown successful suppression of motion artifacts with PHASTE imaging. The image quality was enhanced relative to the original HASTE image, but was still less sharp than a non-motion-corrupted TSE image.

Keywords

HASTE; K-space inherited parallel acquisition; T2 blurring; Carotid artery imaging; Motion suppression

1. Introduction

The half-Fourier single-shot turbo spin-echo (HASTE) sequence [1] is an ultrafast magnetic resonance (MR) imaging technique that acquires slightly more than half of two-dimensional imaging k-space after a single excitation pulse by means of an echo train separated by refocusing pulses. The full k-space is computed from the measured half of k-space data by applying Homodyne detection [2]. The short acquisition time of the HASTE sequence makes it less susceptible to motion, which is one of the main sources of image artifacts in clinical applications [3]. Because of this characteristic, HASTE has been routinely used in abdominal imaging [4–6]. The diagnostic value of HASTE has also been investigated for MR imaging of

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the fetus [7–12] and the brain [13–15]. It has been shown that HASTE-based sequences are particularly useful in imaging uncooperative patients.

However, HASTE has a number of drawbacks. The short acquisition time results in low signal-to-noise ratio (SNR) [5]. The long echo train after the single excitation causes severe T2 blurring artifacts in the phase encoding (PE) direction which can significantly degrade the image quality [14,16]. Because of these drawbacks, HASTE is usually used to image tissues or structures with long T2 relaxation time which are relatively unaffected by T2 blurring, with thick slices and low in-plane resolution to maintain adequate SNR. To make the HASTE sequence applicable for high-resolution imaging of tissues with short T2, it is necessary to shorten the HASTE readout to prevent severe loss of spatial resolution in the PE direction caused by T2 decay. This can be done by either shortening the acquisition time for each echo or reducing the number of echoes acquired, or echo train length (ETL). Gradient hardware limitations and nerve stimulation constraints limit the degree to which the acquisition time for each echo can be reduced. Therefore, the application of partially parallel imaging (PPI) techniques [17,18] to reduce the echo train length for single-shot imaging methods has been extensively investigated.

The PPI technique of simultaneous acquisition of spatial harmonics (SMASH) [6] has been combined with single-shot imaging techniques such as echoplanar imaging (EPI) [19], BURST [20] and HASTE [21]. The images with the SMASH acquisition had increased spatial resolution and improved image quality as compared to conventional single-shot images. However, fixed field of view constraints in coil design for SMASH imaging have limited its wide application. Sensitivity encoding (SENSE) has also been applied to single-shot imaging [22] to reduce the acquisition time. Image reconstruction with partial k-space and parallel imaging has been formulated as an inverse problem with the solution satisfying certain phase constraints [23,24]. The methods mentioned above all require accurate knowledge of coil sensitivity maps. Errors in measurement of coil sensitivities can greatly affect the quality of images reconstructed from these methods. The HASTE sequence has also been combined with generalized auto-calibrating partially parallel acquisition (GRAPPA), which does not require knowledge of coil sensitivities [25,26]. This GRAPPA-HASTE method was applied to improve resolution in lung imaging. Although the lung images were relatively noisy, significant improvement in resolution could be observed in comparison with the images obtained from the conventional HASTE sequence.

Carotid MRI [27,28] has been used for over a decade to study the lumen and wall of the carotid artery with a variety of contrasts such as proton density (PD), T1, and T2 weighting. The turbo-spin-echo (TSE) sequence is most commonly used for carotid MRI. However, the long scan time of a TSE sequence causes artifacts from vessel pulsation, respiratory motion, swallowing or other patient movements. Ghosting artifacts and image blurring due to blood pulsation are the most severe.

In this paper, the application of HASTE to achieve artifact-free T2 weighted carotid artery imaging is investigated. Two limitations of HASTE imaging are addressed to improve its performance for carotid imaging. First, the readout time for a single slice (T_{read}) must be shortened to a fraction of the cardiac cycle. Such a short acquisition time is required for effective suppression of image artifacts caused by blood pulsation. Second, the blurring in HASTE images caused by T2 decay along the long echo train must be reduced. To address these limitations, we chose to combine HASTE with parallel acquisition (PHASTE) and a T2 decay compensation technique. Application of the proposed method to carotid artery imaging has produced T2 weighted images with significantly reduced pulsation artifact and decreased T2 blurring.

2. Theory

The PHASTE sequence acquires only a fraction of the PE views in the half k-space (Fig. 1a) collected by HASTE. Thus, the ETL for PHASTE is reduced, thereby decreasing the total readout time, T_{read} , and also reducing the signal decay due to T2 relaxation. To reconstruct PHASTE data, k-space inherited parallel acquisition (KIPA) [29] was applied to recover the missing PE views (dashed lines in Fig. 1b) in the acquired half k-space. Then, homodyne reconstruction was used to generate the image corresponding to a full k-space (Fig. 1c). For studies with multiple acquisitions, images from all acquisitions were averaged to yield a final image with improved SNR. The combination of HASTE with PPI alone still leaves obvious blurring in the image due to the T2 decay effect. To further reduce the blurring in the PHASTE image, T2 decay compensation was performed before applying KIPA and ramp-weighted homodyne reconstruction.

2.1 T2 Decay Compensation

The need for performing T2 decay compensation is illustrated by a comparison of the energy spectrum of data acquired from a phantom by HASTE and PHASTE sequences with that of data acquired by the TSE sequence (see Fig. 2). The energy spectrum was defined as a function of k_y

$$E(k_y) = \sum_{k_x=1}^{N_x} |S(k_x, k_y)| \quad (1)$$

where $S(k_x, k_y)$ is the signal at a k-space point (k_x, k_y) and N_x is the number of sampling points in the readout direction. The gray curve in Fig. 2a and b is the energy spectrum for TSE data, which were acquired with $ETL=7$ and $N_x = 256$. The peak of the spectrum corresponds to the k-space center. The dark curve in Fig. 2a and b is the energy spectrum for HASTE and PHASTE ($R=2$) data, respectively. Both sequences acquired phase encodings in a linear order, from a small positive value to the most negative value. 8 reference lines were obtained in the low spatial frequency region, leading to an N_x of 136 for HASTE data and 72 for PHASTE data. T2 decay over the long echo train of the HASTE and PHASTE acquisitions causes the peak of the energy spectrum to occur at a k-space line acquired before the center of k-space. T2 decay in HASTE causes attenuation of energy in the high-frequency k-space lines. This effect is lessened by PHASTE, as shown in Fig. 2b.

To reduce the T2 decay effect on the PHASTE image quality, a technique of T2 decay compensation was implemented. In addition to the KIPA reference data, each slice was scanned with the phase encoding gradient off. The T2 decay can be roughly quantified by calculating the average power of each echo in the echo train with no phase encoding:

$$D_j = \sum_{k_x=1}^{N_x} |S(k_x, k_y=0, t=TE_j)| \quad (j=1, \dots, ETL) \quad (2)$$

where j is the reversed index of the echo in the echo train. A plot of D_j with $ETL=136$ is shown in Fig. 3a. The compensation function might be defined as $1/D_j$. However, the inverse of D_j has very large values for high spatial frequencies (Fig. 3b), which would greatly increase the noise level in the reconstructed image. To avoid this noise amplification, it is useful to apodize the compensation function in the high spatial frequency range. Hence, the T2 compensation function was defined using a Gaussian apodization function as:

$$f(k_y^j) = \frac{\max(D_j)}{D_j} (G_\sigma(k_y^j) + \beta) \quad (j=1, \dots, ETL) \quad (3)$$

where G is a Gaussian function with standard deviation σ and centered at $k_y = 0$, and k_y^j is the actual k_y value of the j^{th} echo with j also being the reversed index of the echo in the echo train. A small factor β is used to ensure that the high frequency data are not over-suppressed. The value of the standard deviation reflects a trade off between accuracy of T2 decay compensation and noise amplification. Larger σ corresponds to a wider apodization window and results in a higher noise level. Conversely, smaller σ corresponds to a narrower window, which might over-suppress the high frequencies signal so that noise level is low but the image suffers from resolution loss in the PE direction. From our experience, $\sigma = N_y/6$ gives a good balance between noise level and T2 decay compensation for a Gaussian window with 256 points. The resulted T2 compensation function is shown in Fig. 3d. Finally, the compensated data S^{comp} are calculated by

$$S^{\text{comp}}(k_x, k_y^j) = f(k_y^j) \cdot S(k_x, k_y^j) \quad (j=1, \dots, ETL) \quad (4)$$

2.2 Data acquisition

As the initial step in data acquisition, the HASTE sequence is run to collect a fully sampled half k-space with the phase encoding gradient set to zero to measure the average T2 decay for each echo in the HASTE echo train. Then, a fully sampled HASTE half-k-space acquisition is performed to allow computation of weights for the KIPA parallel imaging technique. Finally, a number of subsampled PHASTE acquisitions are performed, each of which yields a complete image after T2 correction, KIPA processing, and homodyne reconstruction. These images are averaged to give the final image. The number of images acquired and averaged is dictated by SNR requirements.

2.3 Reconstruction of the T2-compensated PHASTE Data

The fully sampled half-k-space HASTE acquisition is corrected for T2 decay, using the T2 correction data obtained with the phase encode gradient set to zero. This T2 compensated fully sampled half-k-space data is then used to compute the KIPA reconstruction coefficients [29]. These coefficients are applied to the T2-compensated PHASTE data to calculate the missing PE views within the half k-space (shown as dashed lines in Fig. 1b). After lines in the subsampled half k-space are estimated, a ramp-weighted Homodyne reconstruction is applied to the half-k-space data to reconstruct an image corresponding to the complete k-space (Fig. 1c) for each coil. The final image is computed as the square root of the sum of squares of the images from individual coils.

3. Materials and Methods

All our experiments were performed on a 3T Siemens Trio scanner (Siemens Medical Solutions, Erlangen, Germany). A standard HASTE sequence was modified to acquire a total number of GRAPPA reference lines equal to the reduction factor R . A reduced number of reference lines (or in fact no reference lines at all) can be used with KIPA since weights are calculated from a fully sampled initial k-space acquisition. These few reference lines eliminate measurement estimation errors at the very center of k-space and thereby reduce ghosting artifacts. TE of 50–60 ms was used, representing a trade-off between higher SNR obtained from shorter echo times and adequate T2 weighting.

A number of phantom and human imaging studies were performed. The detailed protocol and typical results are presented here for a phantom, two normal human volunteers and three human subjects with carotid disease. Informed consent was obtained from the subjects participating in this study in accordance with an institutional review board-approved protocol.

3.1 Phantom Study

A phantom study was performed to compare the performance of conventional HASTE imaging with PHASTE and KIPA reconstruction. A 12-channel head coil (Siemens Medical Solutions, Erlangen, Germany) was used in the study. A phantom was imaged with both the conventional HASTE sequence and the PHASTE sequence. Reconstruction of the PHASTE images was performed with and without T2 compensation of the k-space data before KIPA weight calculations to illustrate the effects of T2 compensation on image resolution. The matrix size for this study was 256×256 , with an ETL of 136 for the HASTE sequence and 72 for the PHASTE sequence.

3.2 Carotid Imaging Studies

Carotid imaging studies were performed on two normal human volunteers and three human subjects with disease in the bifurcation using our custom 4-channel bilateral phased array carotid imaging coil. Informed consent was obtained in all cases. In all studies, measurement data were acquired using the PHASTE sequence. Six axial slices were imaged in each study, with image matrix of 256×256 over a 12.8 cm square field of view and slice thickness of 2–3 mm. The PHASTE sequence was performed with a reduction factor of 2 and ETL of 72, leading to a total readout time of 671 ms and total imaging time of 4 seconds. This scan was repeated four times and averaged, for a total imaging time of 16 seconds. HASTE was used in each study with PE on and off to acquire additional data for the KIPA reference data and T2 compensation. The HASTE sequence had an ETL of 136 with echo spacing of 9.22 ms, giving a total readout time of 1270 ms and total imaging time of 7.5 seconds. Therefore, the additional data acquisition required 15 seconds. Echo times were 74 ms for HASTE and 55 ms for PHASTE.

For comparison, the same slices were also imaged using the TSE sequence. The imaging matrix and spatial resolution were the same as that for the PHASTE sequence. TE was selected to be 62 ms to obtain T2 weighted contrast. The ETL for the TSE scan was 7 and TR was 2000 ms. The scan was interleaved and gated. Only 1 measurement was acquired, resulting in a T_{scan} of 1:53 min. For the human subjects with disease, fat-saturation was applied in the TSE images to reduce artifacts due to skin motion.

3.3 Analysis

To assess relative image quality, TSE and T2 compensated PHASTE images were compared side by side (randomly positioned) to a blinded radiologist observer. A simple forced choice was used to select the “best” image in each of 4 categories: 1) wall sharpness, 2) least motion artifact, 3) best blood suppression, 4) best visualization of the carotid wall and plaque if present.

4. Results

4.1 Phantom Study

The result of the phantom study is shown in Fig. 4. Fig. 4a and b are HASTE and PHASTE images reconstructed without performing the T2 decay compensation. Reduced T2 blurring artifact can be observed in Fig. 4b due to the shorter ETL of PHASTE than that of HASTE. The corresponding T2-compensated HASTE and PHASTE images are displayed in Fig. 4c and d. Although the adopted T2 compensation technique decreases the SNR of the HASTE image,

this SNR loss is partially alleviated for PHASTE data which are obtained from a shorter echo train. The T2-compensated PHASTE image with R=2 (Fig. 4d) is an alias-free, clear image, demonstrating the effectiveness of the proposed method of combing PHASTE with T2 decay compensation.

4.2 Carotid Imaging Studies

The result of the first human carotid study is shown in Fig. 5. Fig. 5a and c are images obtained using the HASTE and TSE sequence, respectively. The TSE image suffers from motion artifacts because of its long acquisition time, causing some ghosting inside the carotid arteries (indicated by the arrows in Fig. 5c). In the HASTE image, the motion artifacts are greatly suppressed but the image shows severe T2 blurring. The blurring has not only degraded the overall image quality, but also distorted the shape of vessels. For example, the left carotid artery (indicated by the arrow in Fig. 5a) is elongated in the PE direction. The PHASTE image as shown in Fig. 5b is greatly improved over the HASTE image. The blurring artifact is alleviated. More importantly, the shape of the left carotid artery has become normal in Fig. 5b. However, the PHASTE image still looks more blurry than the TSE image. Applying T2 compensation to the PHASTE data resulted in the image shown in Fig. 5d, which has strongly reduced the T2 blurring artifact in comparison with Fig. 5b.

Fig. 6 shows the result from the second human carotid study. The three images in Fig. 6a are T2-compensated PHASTE images. The corresponding TSE images are displayed in Fig. 6b for comparison. The motion artifacts present in the TSE images were successfully suppressed in the PHASTE images. Visibility of the carotid arteries in the TSE images is hindered by the ghosting artifacts, whereas in the T2-compensated PHASTE images, the carotid arteries are well visualized (indicated by the arrows in the top two rows). In images which are not severely affected by motion artifact, the TSE sequence give better depiction of the vessel wall than PHASTE. For example, the carotid artery wall in the TSE image in Fig. 6b (indicated by the arrows in the bottom row) is lost in the PHASTE image in Fig. 6a.

Figure 7 shows comparison images from two of the 3 subjects with carotid disease. The results of the observer analysis comparing T2w TSE with T2 compensated PHASTE are shown in Table 1. Significance was determined using the binomial distribution and defining the z-score as [30]:

$$z = \frac{x - np_o}{\sqrt{np_o(1 - p_o)}} \quad (5)$$

Here, p_o , is 0.5, assuming for the null hypothesis that both techniques are equal, and we have $n=18$ separate images. x is the number counted. Similarly, the results of comparing PHASTE without and with T2 compensation are reported in Table 2. Several observations can be made from these two comparisons. First, T2-compensated PHASTE has significantly less motion artifact and better blood suppression than T2w TSE. Further, although there is an indication that T2w TSE provides sharper detail and better plaque visualization, this is not significant in the limited number of studies here. Similarly, comparing PHASTE without and with T2 compensation, it is seen that compensation substantially improves image sharpness and plaque visualization. Interestingly, T2 compensation consistently enhances motion artifacts and any residual signal within the lumen. However, we note again that T2-compensated PHASTE was consistently better than T2w TSE in both of these latter categories.

5. Discussion

Several observations can be made from these two comparisons. First, T2-compensated PHASTE has significantly less motion artifact and better blood suppression than T2w TSE. Further, although there is an indication that T2w TSE provides sharper detail and better plaque visualization, this is not significant in the limited number of studies here. Similarly, comparing PHASTE without and with T2 compensation, it is seen that compensation substantially improves image sharpness and plaque visualization. Interestingly, T2 compensation consistently enhances motion artifacts and any residual signal within the lumen. However, we note again that T2-compensated PHASTE was consistently better than T2w TSE in both of these latter categories.

Although these observations are encouraging, it is still evident that the greatest improvement with PHASTE vs. conventional T2wTSE is observed when the T2w TSE images suffer from motion artifacts. When there is little motion artifact, the TSE images generally show better detail of the carotid wall and plaque. The T2-compensated PHASTE image demonstrates slightly lower resolution than the TSE image. This effect arises from the KIPA reconstruction process, which tends to underestimate high-frequency regions of k-space [29][31,32]. Consequently, the loss of high spatial frequency information causes blurring of the vessel wall. Underestimation of k-space data with KIPA is lessened with increased SNR. A better KIPA reference image might be obtained by a multi-shot acquisition of the half k-space using the TSE sequence. This approach keeps the ETL short and thereby does not cause such high attenuation of the outer k-space signal. It is expected that the motion artifact in this type of reference images will not have significant effect on the reconstruction weights, because the calculation of KIPA weights includes an averaging process that might average out the effect of motion. But further investigation is needed to see if this type of reference image results in better depiction of the carotids walls.

SNR of the high spatial frequency signal of PHASTE data can be improved by further shortening the ETL, which can be achieved by sampling the k-space data with a larger reduction factor. However, the final image SNR will become lower since fewer PE views are acquired. Parallel reconstruction for larger R has additional SNR penalty. Thus the advantages of decreased motion artifact obtained with high reduction factor must be traded off with decreasing SNR and worsening aliasing artifact in the parallel reconstruction.

A promising way to shorten the ETL is to implement inner-volume excitation in the HASTE sequence to reduce the number of phase encoding steps required to obtain a given resolution over the carotid artery. If only magnetization in the region around the carotid arteries is excited, then the number of required PE views can be greatly reduced. Combined with parallel acquisition of R=2, the reduced phase FOV approach may allow a decrease of the ETL to around 20 for resolution equivalent to a full-FOV N_y of 256. Several techniques of inner-volume excitation have been published which could be explored for this application [33].

SNR could be improved by acquiring multiple averages, improved RF coil design, or optimizing the PHASTE sequence to conserve more signal energy along the echo train.

The scan time needed for acquisition of the T2 compensation data can be shortened considerably. We have found that average T2 decay was similar for different slices from carotid artery imaging. Thus, acquisition of the HASTE data without PE for T2 compensation is only needed for one slice from the imaged slice group.

We have applied PHASTE to carotid imaging, where motion artifact from vascular pulsation limits the usefulness of conventional TSE imaging. Using PHASTE, k-space data for a single slice can be acquired within a fraction of the cardiac cycle. Thus, T2-compensated PHASTE

with the KIPA plus Homodyne reconstruction method might be an appropriate technique for other applications that need very fast imaging to suppress cardiac motion.

6. Conclusion

In this paper, we presented some preliminary results from our studies on reducing motion artifacts for T2 weighted carotid artery imaging. The motion artifacts due to blood pulsation were suppressed to a negligible level by acquiring data using the very fast PHASTE sequence. A simple T2 compensation technique was applied to the PHASTE data to reduce the T2 decay artifact. The T2-compensated PHASTE data were reconstructed using the KIPA algorithm followed by a ramp-weighted homodyne reconstruction to produce images with negligible aliasing artifact. However, the quality of the carotid images acquired with PHASTE is still not equal to that of a non-motion-corrupted TSE image. To obtain clinically useful carotid images, further development of the PHASTE carotid imaging method may be required.

Acknowledgements

This work is supported in part by NIH grants R01 HL57990, Siemens Medical Solutions, the Cumming Foundation, and the Ben B. and Iris M. Margolis Foundation.

References

1. Nozaki A. Single shot fast spin echo (SSFSE). *Nippon Rinsho* 1998;56(11):2792–7. [PubMed: 9847599]
2. Noll DC, Nishimura DG, Macovski A. Homodyne detection in magnetic resonance imaging. *IEEE Trans Med Imaging* 1991;10(2):154–63. [PubMed: 18222812]
3. Terk MR, et al. Halfscan: clinical applications in MR imaging. *Magn Reson Imaging* 1991;9(4):477–83. [PubMed: 1779717]
4. Mirowitz SA, et al. Rapid acquisition spin-echo (RASE) MR imaging: a new technique for reduction of artifacts and acquisition time. *Radiology* 1990;175(1):131–5. [PubMed: 2315472]
5. Semelka RC, et al. HASTE MR imaging: description of technique and preliminary results in the abdomen. *J Magn Reson Imaging* 1996;6(4):698–9. [PubMed: 8835965]
6. Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays. *Magn Reson Med* 1997;38(4):591–603. [PubMed: 9324327]
7. Levine D, et al. Fetal anatomy revealed with fast MR sequences. *AJR Am J Roentgenol* 1996;167(4):905–8. [PubMed: 8819380]
8. Tsuchiya K, et al. Short communication: MR imaging of fetal brain abnormalities using a HASTE sequence. *Br J Radiol* 1996;69(823):668–70. [PubMed: 8696706]
9. Yamashita Y, et al. MR imaging of the fetus by a HASTE sequence. *AJR Am J Roentgenol* 1997;168(2):513–9. [PubMed: 9016238]
10. Levine D, et al. Fetal fast MR imaging: reproducibility, technical quality, and conspicuity of anatomy. *Radiology* 1998;206(2):549–54. [PubMed: 9457211]
11. Huppert BJ, et al. Single-shot fast spin-echo MR imaging of the fetus: a pictorial essay. *Radiographics* 1999;(19 Spec No):S215–27. [PubMed: 10517456]
12. Kubik-Huch RA, et al. Ultrafast MR imaging of the fetus. *AJR Am J Roentgenol* 2000;174(6):1599–606. [PubMed: 10845491]
13. Patel MR, et al. Half-fourier acquisition single-shot turbo spin-echo (HASTE) MR: comparison with fast spin-echo MR in diseases of the brain. *AJNR Am J Neuroradiol* 1997;18(9):1635–40. [PubMed: 9367310]
14. Sugahara T, et al. Comparison of HASTE and segmented-HASTE sequences with a T2-weighted fast spin-echo sequence in the screening evaluation of the brain. *AJR Am J Roentgenol* 1997;169(5):1401–10. [PubMed: 9353469]
15. Penzkofer AK, et al. MR imaging of the brain in pediatric patients: diagnostic value of HASTE sequences. *AJR Am J Roentgenol* 2002;179(2):509–14. [PubMed: 12130464]

16. Constable RT, Gore JC. The loss of small objects in variable TE imaging: implications for FSE, RARE, and EPI. *Magn Reson Med* 1992;28(1):9–24. [PubMed: 1435225]
17. Blaimer M, et al. SMASH, SENSE, PILS, GRAPPA: how to choose the optimal method. *Top Magn Reson Imaging* 2004;15(4):223–36. [PubMed: 15548953]
18. Wilson GJ, et al. Parallel imaging in MR angiography. *Top Magn Reson Imaging* 2004;15(3):169–85. [PubMed: 15479999]
19. Mansfield P. Multi-planar image formation using NMR spin echoes. *J Phys C Solid State* 1977;10:55–8.
20. Hennig J, Mueri M. Fast imaging using burst excitation pulses. *MAGMA* 1993;1:39–48.
21. Griswold MA, et al. Resolution enhancement in single-shot imaging using simultaneous acquisition of spatial harmonics (SMASH). *Magn Reson Med* 1999;41(6):1236–45. [PubMed: 10371457]
22. Pruessmann KP, et al. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med* 1999;42(5):952–62. [PubMed: 10542355]
23. Willig-Onwuachi, JD., et al. Phase-constrained parallel MR image reconstruction using symmetry to increase acceleration and improve image quality; the 11th Annual Meeting of ISMRM; Toronto, Canada. 2003.
24. Bydder M, Robson MD. Partial fourier partially parallel imaging. *Magn Reson Med* 2005;53(6):1393–401. [PubMed: 15906299]
25. Heidemann RM, et al. Resolution enhancement in lung 1H imaging using parallel imaging methods. *Magn Reson Med* 2003;49(2):391–4. [PubMed: 12541262]
26. Griswold MA, et al. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med* 2002;47(6):1202–10. [PubMed: 12111967]
27. Choudhury RP, et al. MRI and characterization of atherosclerotic plaque: emerging applications and molecular imaging. *Arterioscler Thromb Vasc Biol* 2002;22(7):1065–74. [PubMed: 12117718]
28. Rapp JH, Saloner D. Current status of carotid imaging by MRA. *Cardiovasc Surg* 2003;11(6):445–7. [PubMed: 14627964]
29. Guo JY, et al. K-space Inherited Parallel Acquisition (KIPA): application on dynamic magnetic resonance imaging thermometry. *Magn Reson Imaging* 2006;24(7):903–15. [PubMed: 16916708]
30. Zar, JH. *Biostatistical Analysis*. 3. Upper Saddle River, New Jersey: Prentice Hall; 1996.
31. Maier SE. Slab scan diffusion imaging. *Magn Reson Med* 2001;46(6):1136–43. [PubMed: 11746580]
32. Pauly J, Nishimura D, Macovski A. A k-space analysis of small-tip-angle excitation. *J Magn Reson* 1989;81:43–56.
33. Jeong EK, et al. High-resolution DTI with 2D interleaved multislice reduced FOV single-shot diffusion-weighted EPI (2D ss-rFOV-DWEPI). *Magn Reson Med* 2005;54(6):1575–9. [PubMed: 16254946]

Fig. 1. Illustration of the PHASTE data reconstruction where the denser sampling is around the center of k-space. (a) The k-space sampling pattern for PHASTE with $R=2$. (b) The k-space data after KIPA was used to recover the missing PE views (dashed lines). (c) The complete k-space corresponding to the image obtained after applying Homodyne reconstruction to the data in (b).

Fig. 2. Comparison of energy spectrum for different pulse sequences. (a) Energy spectrum for TSE (gray line) and HASTE (dark line). (b) Energy spectrum for TSE and PHASTE with R=2 (dark line).

Fig. 3.

Illustration of T2 compensation function construction. (a) Function D calculated from the data without PE. (b) Inversion of function D with the minimum normalized to 1. (c) The sum of Gaussian function ($\sigma=128/3$) with a factor $\beta=0.01$. (d) The final T2 compensation function obtained by multiplying the functions shown in (b) and (c).

Fig. 4. Phantom study. (a) HASTE image without T2 compensation. (b) PHASTE image without T2 compensation, $R=2$. (c) HASTE image with T2 compensation. (d) PHASTE image with T2 compensation, $R=2$.

Fig. 5. Result of the first human carotid study. (a) HASTE image without T2 compensation. (b) PHASTE image without T2 compensation. (c) TSE image. (d) PHASTE image with T2 compensation.

Fig. 6. Result of the second human carotid study. (a) T2-compensated PHASTE images for different slices with different slice thickness (3 mm for the top 2 images and 2 mm for the bottom image). (b) TSE images for the same slices.

Fig. 7. Comparison of T2 weighted TSE images (left column) with T2-compensated PHASTE images (right column). Top two rows are from a single subject with minimal motion artifact. Bottom row is from a subject with substantial motion artifact. Thin arrows point to carotid artery with plaque. Fat arrows point to carotid artery obscured by the motion artifact.

Table 1

Forced choice comparison of conventional TSE with T2 compensated PHASTE in terms of which image shows best wall sharpness, least motion ghosting, greatest suppression of blood signal in the lumen, and best visibility of the wall, including plaque in the wall

Measure	TSE	T2PH	z-score	p-value
Wall Sharpness	12	6	1.414	p>0.1
Motion Ghosts	0	18	4.24	p<0.001
Black Blood	2	16	3.30	p<0.005
Plaque	10	8	0.47	p>0.5

Table 2

Forced choice comparison of PHASTE without and with T2 compensation in terms of which image shows best wall sharpness, least motion ghosting, greatest suppression of blood signal in the lumen, and best visibility of the wall, including plaque in the wall

Measure	PH	T2PH	z-score	p-value
Wall Sharpness	0	18	4.24	p<0.001
Motion Ghosts	18	0	4.24	p<0.001
Black Blood	14	4	2.35	p<0.05
Plaque	1	17	3.77	p>0.002