A REAL ASYMMETRIC FOURIER IMAGING (REAL-AFI) WITH PRESERVING PHASE POLARITY FOR INVERSION-PREPARED SPIN-ECHO BASED-SEQUENCE

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Target Audience: Researchers and clinicians who are interested in MRI reconstruction technique

Purpose: A phase-sensitive (real-part) imaging combining with inversion-prepared (IR) sequence is useful for T1W[1], FLAIR[2], and black-blood (BB) imaging [3]. An asymmetric Fourier Imaging (AFI), which is a MR image reconstruction technique combining phase correction and Hermitian conjugate theory, is also widely used to reduce imaging time or TE while minimizing blur especially in fast-spin echo (FSE) -based sequence, because the background phase errors are smaller than in the GRE-based sequence. Several AFI techniques have proposed and applied [4-5]. However they commonly cannot provide phase information because phase correction is performed using symmetric sampled low-frequency k-space region before conjugation-equivalent process; accordingly, it is not used for IR sequence when positive and negative signals were mixed. The purpose of this study was to propose and assess a modified AFI technique called RealAFI where the phase polarities can be preserved when applying to IR-FSE truncated k-space data for BB.

Theory: When the inversion time (TI) of IR imaging is shorter than that of nulling Mz (TI_{null}), positive and negative signals are mixed and there is no phase between 0 and |π| at an ideal condition. However, actual MR signals include additional background phase Φ_{back}. An ideal MR signal is regarded as vector summation of complex conjugate pair signals V^+ and V^- ; and in AFI, the unknown V^- (in no sampled k-space region) is estimated from the known V^+ (sampled k-space region). If Φ_{back} is estimated from the self data, V_{ideal} can be obtained by twice of real part of V^+ (**Fig. 1**). Here we consider a case for truncated in 1D direction to simplify and define as follows: k-space data is S(k), real-space (r-space) data is V(r), and AFI image data is $I_{cor}(r)$; original partially sampled k-space data: $S_{orig}(k)$: $-K_c<=k<=K_{max}$ ($K_c<K_{max}$), where zeros are filled in un sampled region ($-K_{max}<=k<K_c$); and several window functions are:

a) *circular symmetric low-pass filter* for back-ground phase estimation: $H(k_r)=\exp[(-\ln 2)(k_r/K_{r2})^2]$: $0<|k_r|<=K_c$, =0: otherwise, where k_r is radial in polar coordinate in k-space in 2D or 3D data..

b) *homodyne high-pass filter*: $H_{high.homo}(k)=H_{low.homo}(k)$: $k<0$, $=2-H_{low.homo}(k)$: $k>=0$; where $H_{low.homo}(k)=\exp[(-\ln 2)\{(k-(K_c-K_1))/K_2\}^2]$: $K_c-K_1<|k|<=K_c$, =0: otherwise;

c) *merge filter*: $H_{whole}(k)=H_{low.homo}(k)$: $-K_{max}<=k<0$, =1: otherwise; *Conj*[ ] means taking complex conjugate. The RealAFI algorithm assessed here was generated by modifying the process of low-pass filtering in standard POCS algorithm [5]:

1) Low-pass filtering: $V_{low}(r)=FT[H_{low}(k_r)*S_{orig}(k_r)]$.
2) Homodyne filtering: $V_{high.homo}(r)=FT[H_{high.homo}(k)*S_{orig}(k)]$.
3) Phase correction: $V_{cor}[r]=V_{high.homo}[r]*Conj[V_{low}(r)]/|V_{low}(r)|$.
4) Take real-part in r-space: $I_{cor}(r)=Real[V_{cor}(r)]$.

The next step of 5) to 8) is iterated for N times (n=1~N) until convergence.

5) Phase restoration: $S_{cor}(k,n)=IFT[I_{cor}(r,n-1)*V_{low}(r)/|V_{low}(r)|]$.
6) $S_{merge}(k,n)=\{1-H_{whole}(k)\}*S_{cor}(k,n)+H_{whole}(k)*S_{orig}(k)$.
7) phase-correction: $V_{cor}(r,n)=FT[S_{merge}(k,n)]*Conj[V_{low}(r)]/|V_{low}(r)|$.
8) Take real-part in r-space : $I_{cor}(r,n)=Real[V_{cor}(r,n)]$.

Methods:

***Simulations*:** Simulations were performed assuming 1D vessel profile, 2nd order background phase of $\Phi_{back}=0.0002*k^2$ [rad], and internal vessel phase of 180 degree (ideally inverted case) and shifted cases from the ideal. 3 size of vessels width=10, 8, 5 were assumed. $K_{max}=128$, $K_c=16$, $K_{r2}=4$, $K_1=8$, and $K_2=4$ were used.

***MR Experiments*:** MR Experiments were performed using double IR (DIR)-BB-FSE sequence [6] selecting TI was shorter than the TI_{null} of blood. Normal volunteer neck axial data was acquired on Toshiba Vantage Titan™ 3T (Otawara, Japan) after obtaining written informed consent. Parameters were: ETL=8, TR/TE/TI=13000ms/10ms/40ms, slice thickness of 5mm, FOV=20cm, acquisition matrix=224(PE)x384(RO) (fully sampled), display matrix=384x383 (pixel size=0.5x0.5mm²) after sinc interpolation, no cardiac gating, and parallel imaging of reduction-factor, R=2. For AFI parameters, K_c=8 to 64, truncated the front of phase-encode direction (anterior-posterior), $K_{r2}=4$, $K_1=8$, and $K_2=4$, N=4 were used.

***Evaluation*:** RealAFI and real-IR made from fully-sampled data were compared visually and quantitatively using Root Mean Square Error (RMSE) in whole pixels.

Results and Discussion:

In simulation (**Fig. 2, Table 1**), errors (RMSE) in the real-part profiles were increased with increasing vessel phase difference from 180 degree. Note that the errors in 180 degree case were negligible, though the measured phase introduced slightly larger errors than the ideal did. In DIR-BB-FSE imaging (**Fig. 3**), RealAFI provided similar images as the real images obtained from fully sampled data when K_c was selected greater than the twice of low-pass filter size ($\sim 2*K_{r2}$) for background phase estimation. It may be due to motion (flow) induced phase after inversion, because there were potions of phase below |π| in blood vessels even at the fully sampled data even though the background phase was almost correctly estimated by 2D low-pass filter instead of 1D.

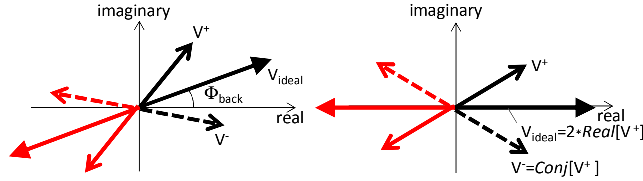
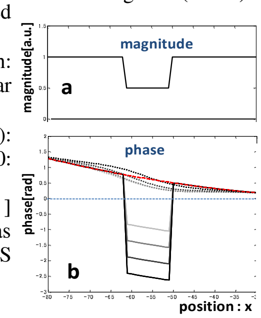


Fig. 1 Principle of real AFI for IR signals with polarity.
This shows vectors before (left) and after (right) background phase correction in complex space. Black and red are respectively positive (Mz>0) and negative (Mz<0) signals.

Table 1 RMSE for real profiles corrected by background phase of ideal and measured

Vessel Phase	90	120	150	180
ideal	1.17	1.01	0.59	0.11
measured	1.61	1.51	0.95	0.55



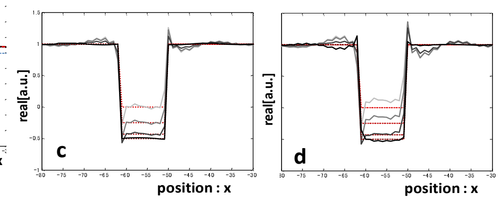
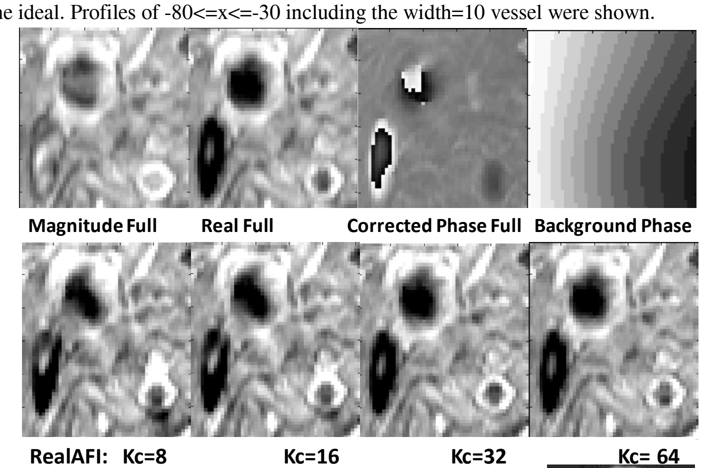


Fig. 2 1D simulated profiles dependency on vessel phase.
magnitude (**a**) and phase (**b**) for original (vessel phase of 90,120,150,180°(light to dark) and 2nd order ideal (red) and measured background phase (dotted); real-part profiles corrected by background phase of ideal (**c**) and measured (**d**), where dotted red lines were the ideal. Profiles of -80<=x<=-30 including the width=10 vessel were shown.



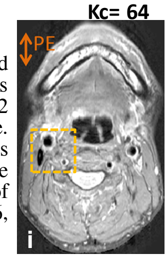


Fig. 3 Results for DIR-FSE PDWI neck axial data.
a-c are made from fully sampled (k=-112~112) data, and **d** is low-passed phase map and the bottom row is images of using partial data of several K_c. It was required K_c>=32 to preserve the vessel shape of fully sampled real image. The background phase obtained here in this area (**d**) was distributed in the range of -2.5~-1.2 [rad]. Whole image and the magnified area are shown in **i** . RMSEs of RealAFI between the Real Full for whole FOV were 596, 486, 349, and 220 (K_c=8, 16, 32, and 64).

Conclusion: Our proposed RealAFI is practically useful technique from the views of balancing image quality and simplicity for IR-FSE sequences, since it allows only by modifying the low-pass filter for background phase estimation in standard AFI technique. Although further intelligent phase estimation will be required when inverted regions are relatively larger than the non-inverted region, it is considered to be applicable in almost cases for FSE-based IR- or DIR-BB imaging.

References: [1] Hou P *et al. AJNR* 26:1432–1438 (2005); [2] Kimura et al. Proc. of 19th ISMRM,pp2649 (2011); [3] Wang et al. MRM69:337-345 (2013); [4] Macgibney et al. MRM 39:51-59(1993); [5] Haacke, JMR, 92,126 (1991); [6] Edelman et al. Radiology181:655–660 (1991)