Point Spread Function Artefacts from In-plane Constant and Pulsatile Flow in EPI: Implications for Pulsatility Imaging

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Introduction New data obtained by rapid (TR=0.1s) gradient echo EPI of normal subjects and flow phantoms show remarkably good low resolution detection of pulsatile flow in vascular structures in the head and unexpectedly faithful flow velocity waveforms [1]. However, the method as it stands suffers from three types of artefact:

(i) within straight vessels where smooth flow is expected, waveforms at different locations can be strongly modulated in amplitude;
(ii) at some locations within the vessel the flow waveform is can be spuriously inverted;
(iii) ghost artefacts of both temporal phases arise outside the vessel.

In this work we present the results of a simulation and actual image results using both simple and physiological flow phantoms. This work is an extension of the early work of Butts and Riederer [2] for the case in in-plane constant and pulsatile flow.

Method
1. *Computer simulation*. A simulated flow phantom consisting of a straight tube with a 75% stenosis and parallel return path was generated on a 32x32 matrix in Matlab 5.2. Flow was orientated either parallel to the frequency encode or phase encode axis and described by a constant velocity term and n harmonics

$$V = V_c + \sum A_n \sin(2\pi f_n t)$$

In the data acquisition, each k-space component is modified by a phase factor

$$\Delta\phi = 2\pi\gamma \int \mathbf{G}(t)\,.\mathbf{r}(t)\,dt$$

where G is the time dependent gradient waveform and $\mathbf{r}(t)$ the instantaneous positional vector of the moving spins. Following inverse 2d FFT of the image of the static phantom we compute point-by-point the accumulated phase errors for a Cartesian "blipped" single shot k-space trajectory. We also note, and make use of the partial velocity compensation on even echoes from the bipolar frequency encode gradient. Thus when flow is along k_X, consecutive phase increments are added for odd lines but subtracted for even lines, and for flow along k_Y, the phase is assumed to accumulate blip by blip at the end of each line of k_X. We assume a Cartesian k-trajectory from $(-k_{Xmax}, -k_{Ymax})$ to $(+k_{Xmax}, +k_{Ymax})$. The final simulated image is generated by 2d FFT of the modified (k_X,k_Y) array. A scale factor α is defined

$$\alpha = V\Delta t\, N^2 / FOV$$

where Δt is the data sampling interval and N the matrix dimension giving a total acquisition time $\Delta t\, N^2$.

2. *Phantom experiments* Scanning was performed on a Siemens Vision 1.5T MRI using single shot blipped EPI with half sinusoidal frequency encode gradients on a 64x64 matrix with TE= 22ms, and also on a 96x128 matrix with TE =40ms. In both instances a TR of 0.1s was used, and data collected continuously for 10s using a single 10mm slice arranged such that the phantom produced in-plane flow. A programmable UHCD pulsatile carotid phantom [3] was imaged in addition to a simple gravitational constant flow straight phantom. For the carotid phantom T1 was of the order of 70ms, and of the order of 100ms for the simple phantom to avoid inflow effects.

Results Point spread functions for constant flow in each direction are shown in figure 1 for α = 0.1,0.3 and 0.6 or increasing velocity. As expected the flow parallel to frequency encode produces significant out-of-vessel ghosts arising from the even-odd echo phase effect. Flow along phase encode results in much less ghosting. In actual phantom images (Figure 2), the superposition of the PSF results in intensity modulations along the vessel. The appearance of ghosts in the phantom images agrees qualitatively with the simulations. Additionally, the model correctly predicts the occurrence of temporal phase inversions in the time series when the TR is less than 0.5s.

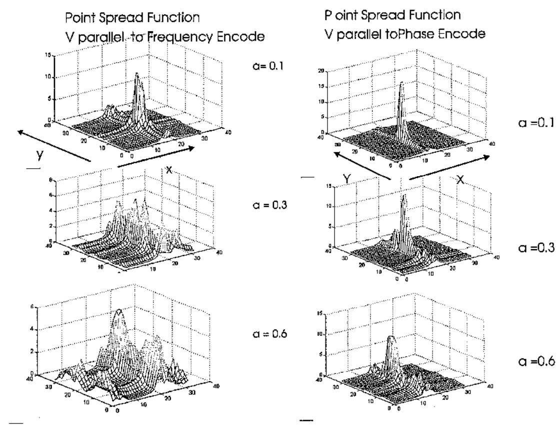

Figure 1 Simulated PSFs for flow parallel to FE and PE axes.

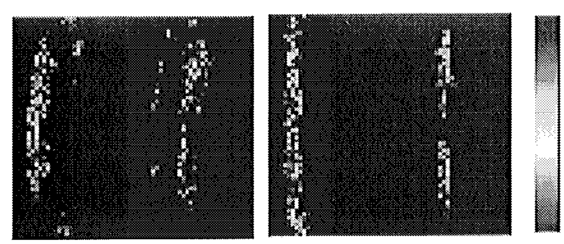

Figure 2 Amplitude data for carotid phantom (stenosis on the RHS): left - flow along FE axis; right - flow along PE axis.

Conclusions These artefacts will affect all EPI acquisitions to a degree that depends upon the velocity, directionality, FOV and sampling times. The clinical development of direct pulsatility EPI [1] will depend upon their removal or minimisation. At the longer TRs used in fMRI, temporally aliassed ghost flow signals may contribute a significant source of noise [4].

References
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