On the battle between Rician noise and phase-interferences in DWI

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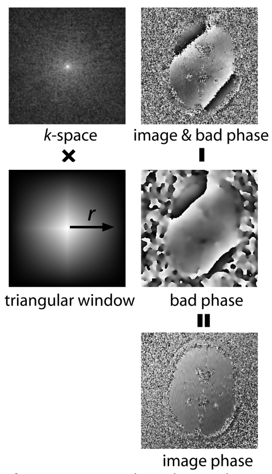

Figure 1. Triangular phase correction [6] applied to each DW k-space in EPI. A window radius of width *r* is used to remove unwanted phase due to motion occurring during the DW gradients.

Introduction: It has been shown that complex averaging benefits diffusion-weighted imaging (DWI) by reducing the influence of the noise bias in the ADC and DTI (1). Magnitude averaging in DWI is traditionally used to combine multi-directional and multi-NEX diffusion data – to avoid the complication of random phase offsets, due to motion that occurs during the DW gradients. However, magnitude averaging introduces a non-zero bias of the signal expectation value (a.k.a. Rician noise (2-4)) and underestimates the calculated ADC (1,5). This bias grows with lower SNR in each diffusion weighted image – and this is a particular concern for high-resolution, thin-slice DWI/DTI. Even in the clinical practice – where the non-quantitative mean DWI (a.k.a. isotropic DWI) is of most importance – the hazy looking image can confound image interpretation and diagnostic confidence.

The advent of phase correction techniques – such as the triangular windowing approach initially proposed for PROPELLER DWI data [6] – has been shown to be a fast and effective phase correction approach for navigated DW EPI data. Here, each complex DW *k*-space is reconstructed twice into two temporary images, one of them after the application of a triangular window function of radius *r* in *k*-space. The non-windowed image is then subtracted with the phase information content in the corresponding windowed image (Fig. 1). The radius of the triangular window will determine how much phase is removed. While a large *r* will help to remove image artifacts due to pulsatile brain motion that occurs during the DW gradients, it will also 'approach magnitude averaging' or - in other words – result in pronounced Rician noise in the mean DWI data after averaging the multiple NEX, repetitions, and diffusion directions together.

In this abstract, we report on the best choice of triangular windowing radius for use in clinical practice. Over 1,200 patients have been scanned at our hospital with our GRAPPA-accelerated EPI sequence [7] and online reconstruction developed in-house. With this data, we have determined the most suitable triangular window radius that minimizes the amount of Rician noise in the final iso-DWI data, without introducing phase cancellations.

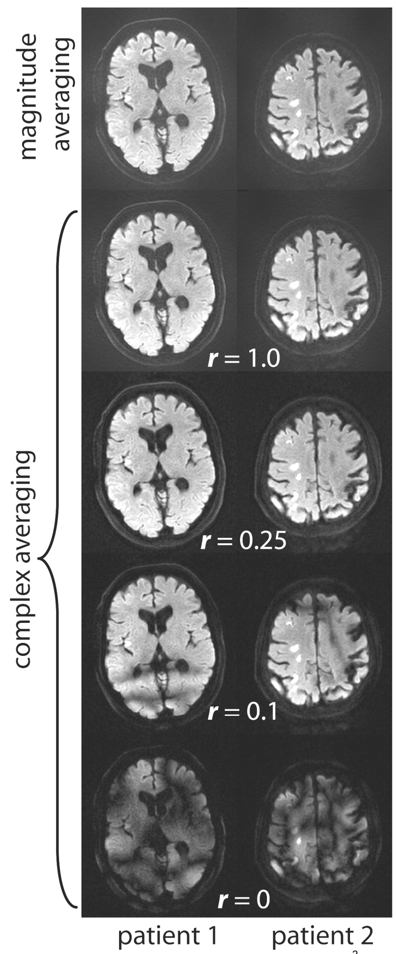

Figure 2. Isotropic DW images (b = 1000 s/mm^2) of two stroke patients reconstructed with magnitude averaging, as well as complex averaging for various triangular-window radii.

Materials and Methods: Our GRAPPA-accelerated EPI pulse sequence has been scanned on a range of 1.5T whole-body GE systems (Waukesha, WI, USA), primarily using an 8-channel head coil. Relevant imaging parameters include: matrix size 192×192; FOV = 24 cm; ~23 slices; slice thickness = 5 mm; TR/TE = 3000/70 ms; a GRAPPA [8,9] factor R = 3; NEX = 9; twice refocusing diffusion preparation; tetrahedral encoding with b = 1000 s/mm^2; scan time = 2:21 min. DWI images from a selection of stroke patients were phase corrected using the triangular-windowing approach, with the use of the following window radii (r): 1, 0.5, 0.4, 0.3, 0.25, 0.2, 0.1, and 0 (the latter represents complex averaging *without* phase correction). One of the datasets had pronounced through- and in-plane motion. For reference, magnitude reconstruction was performed also, by using the absolute value of each image. In all cases, the resulting images were combined over coils with a phase preserving sum-of-squares operation, and averaged over diffusion directions. Our phase preserving sum-of-squares formula is defined as

$$I_{cplxSOS} = \underbrace{\left(\frac{\sqrt{\sum \left|I_j\right|^2}}{N_{coils}}\right)}_{\text{magn. part}} \underbrace{\frac{\sum I_j}{\left|\sum I_j\right|}}_{\text{phase part}} \quad ; \mathbf{I} = [I_1, ..., I_{Ncoils}] \qquad [1]$$

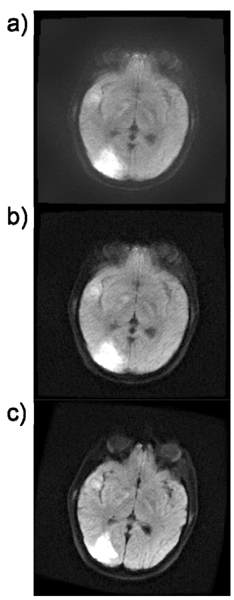

Figure 3. A clinical DWI exam corrupted by motion. The data is reconstructed with (c) and without (a,b) motion correction, as well as magnitude aver-aging (a) and complex aver-aging (b,c) with *r*=0.25.

Results: Fig. 2 shows isotropic DWI images of two stroke patients, reconstructed with both magnitude and complex averaging – the latter with various r. With decreasing r, one can observe decreasing image 'haze'. However, for $r \leq 0.1$, phase cancellation artifacts significantly hamper the image quality. Moreover, artifacts were visible in approximately 5% of the slices reconstructed with r = 0.2. Thus, for routine use at the hospital, the best choice of r was found to be 0.25. Fig. 3 shows data from a patient with head motion, processed with both magnitude averaging and complex averaging using r = 0.25. Fig 3a shows the magnitude averaged data - without motion correction. Fig. 3b and 3c, have been reconstructed without and with motion correction with complex averaging using an r = 0.25. This suggests that this radius is of no concern, even in the presence of head motion.

Discussion: As shown in Figs. 2 and 3, DWI benefits considerably from the reduced background signal in complex averaging. According to [5], for b-values ranging from 300 to 1500 s/mm^2, the magnitude image can give rise to a variation in calculated ADC value of approximately 30%: while variation in the complex reconstruction varies approximately by only 5%. However, as shown in Fig. 2, even more effect can be achieved by reducing r – as lower spatial frequencies of the phase are also preserved, leading to a better noise cancellation when averaged. It could be argued that cardiac gating (which is not currently used in most clinical practices) could potentially allow for a further reduction in this value. This needs to be investigated – although preliminary findings indicate that these phase cancellation artifacts also occur at the top of the brain, where brain motion is not as significant.

References: [1] Dietrich MRM 2001;45:448-453. [2] Gudbjartsson MRM 1995;34:910-915. [3] Henkelman Med Phys 1985;12(2):232-233. [4] Miller MRI 1993;11:1051-1056. [5] Newbould ISMRM 2008; 1810. [6] Pipe MRM 2002;47(1):42-52. [7] Skare MRM 2007;57(5):881-890. [8] Griswold MRM 2002;47:1202-1210. [9] Qu JMR 2005;174(1):60-67.

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