GRAPPA-accelerated dual-echo diffusion-weighted EPI with intensity correction

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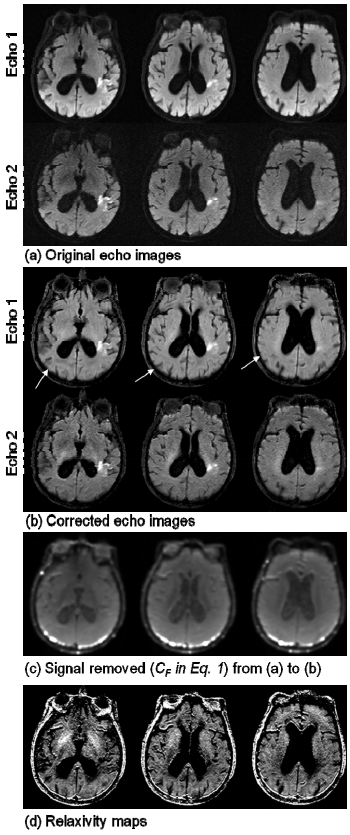

Fig. 1: Data collected on a 91 yr old male stroke patient (a) Original dual echo images, and (b) the dual echo images with contribution from coil sensitivity, proton density, initial magnetization and B1 removed (as in (c)). The corrected images rule out any suspicion of posterior lesions (white arrows). Note the increased contrast of the stroke lesions on echo 2. (d) Relaxivity maps calculated from the average contribution of the T2-w and DWI dual-echo images.

Introduction: A well-known problem with standard single-shot (ss)-EPI for diffusion-weighted imaging (DWI) is geometric distortion and blurring. These artifacts can be reduced through the use of parallel imaging. The shortened readout also affords a shorter echo time - which reduces T2-shine through, but may result in reduced lesion conspicuity on DWI. Thus, in an attempt to both improve the image quality and retain the sensitivity of lesions through longer echo times, here we implement a GRAPPA-accelerated[1,2], dual echo[3] EPI DWI sequence on stroke patients and present our preliminary data. For cases of absent coil sensitivity calibration scans, we also demonstrate how the two echoes can be utilized to remove array coil-induced intensity modulations, which are often confused as potential ischemically challenged regions.

Methods: Patient data was collected on 15 patients suspected of stroke using a 1.5T GE system and an 8-channel head coil. The following scan parameters were used: Stejskal Tanner diffusion preparation with tetrahedral encoding (b = 1000s/mm^2) and one b=0, matrix size = 192 x 192, GRAPPA acceleration factor R = 3 and 3 interleaves (used for the ghost- and GRAPPA-weights estimation as well as to boost SNR), NEX = 2, TR = 3s, TE_1/TE_2 = 51ms/115ms, FOV = 24cm, slthck/gap = 5 mm/1.5mm, and a scan time of 2:15min. The first and second echo were corrected for signal intensity variation across the images using the following:

$$Dc_i = \frac{D_i}{C_F} \quad \text{for } i = 1,2 \qquad \text{Eq. 1}$$

where Dc_i represents each corrected echo image, D_i are the original echo images, C_F is the Gaussian-filtered contribution of, C, given by:

$$C = \sum_{i=1}^{2} D_i \frac{e^{-t_i K}}{\left(\frac{D_1}{I_1}+\frac{D_2}{I_2}\right)} \quad \text{where} \quad K = \frac{1}{2(t_2-t_1)}\left(\log\left(\frac{I_2}{I_1}\right)+\log\left(\frac{D_2}{D_1}\right)\right) \qquad \text{Eq. 2}$$

and C represents contributions to the signal intensity from coil sensitivity, proton density, original magnetization, and RF; and for the two echoes: t = TE, I are the b=0 images, D are the DWI images.

Results: Fig. 1-2 shows GRAPPA-accelerated dual-echo EPI patient data acquired on two patients with strokes of the middle cerebral artery (MCA). The increased lesion conspicuity of echo 2 is apparent. Fig. 2b shows coil sensitivity-corrected maps, which remove the hyperintensity of the signal - particularly in the posterior regions where cortical regions are closer to individual coil elements. Fig. 2 shows patient data where the second echo on isoDWI provided increased diagnostic confidence. Fig. 2b shows isotropic ADC maps calculated from echo 1 and 2, suggesting that the first echo should always be used for its higher SNR.

Discussion: Here we show that GRAPPA-accelerated DWI can be made even more applicable in a clinical setting with the acquisition of a second echo in the same TR. While the first echo can be used for high SNR ADC maps, the second echo can be used for lesion detection (Fig. 1-2). The use of the second echo for lesion detection goes against the common teaching that TE should be kept short to reduce T2-shine through. While this might be true for differentiating between acute and subacute lesions, for general lesion conspicuity longer echo times may yield greater diagnostic confidence. Interestingly, on long-TE DWIs, the combined effect of diffusion restriction and prolonged T2 gives additional contrast that is not necessarily observed on long TE T2-w FSE scans or FLAIR alone. The added information from quantitative ADC and T2 will further the ability to differentiate between acute and subacute lesions. In addition, the two echoes can be used to remove the contribution of coil sensitivity which can lead to misdiagnosis (since hyperintensity around the cortex on DWI can indicate pathology such as stroke, encephalitis, Creutzfeldt-Jakob disease, and epilepsy). This bias field removal may also improve automated segmentation procedures such as used for diffusion-perfusion-mismatch calculation[4]. One could argue that a calibration scan could be acquired for coil-sensitivity correction - however this comes at an additional scan time cost. With the set of imaging parameters that we routinely use for our stroke protocol at our institution, the dual-echo DW-EPI approach does not increase the scan time since the second echo fills in the dead-time of the sequence.

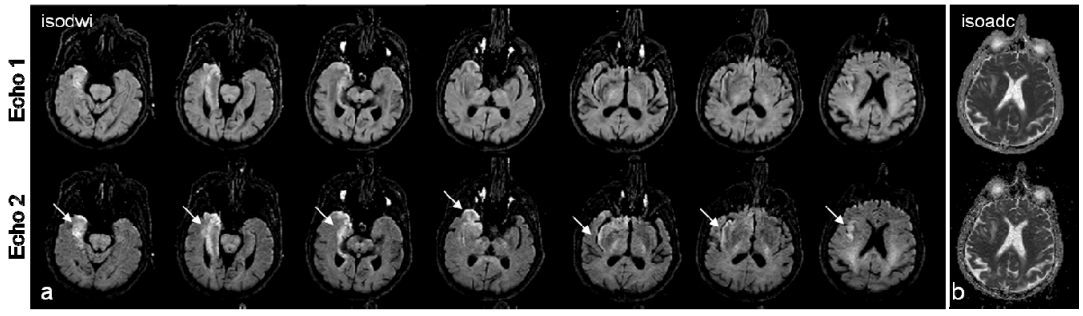

Fig. 2 - Patient data acquired on an 88yr old male stroke patient. (a) Coil sensitivity-corrected isotropic DWI (b=1000s/mm^2) dual echo images showing increased lesion conspicuity on the second echo (white arrows). (b) Single slice from the same dataset showing the isotropic ADC calculated from the b=0 and DWI images from echo 1 (top) and echo 2 (bottom), respectively. The isoDWI echo 2 may be useful for lesion conspicuity, while isoADC echo 1 could be used for its higher SNR.

Conclusion: Here we show that by observing DWI images acquired at two echo times, complementary information can be gleaned, at no additional scan time cost. In addition, the two echoes can be used to remove the coil sensitivity contribution to the DWI images, which may provide clinical confidence and will improve the performance of automated segmentation procedures used to suggest treatment outcome for stroke patients.

References: [1] Griswold, M. *et al*. MRM 2002;47:1202-1210. [2] Qu, P. *et al*. JMR 2005;174(1):60-67. [3] Feinberg, D. et al. MRM 1994:31:461. [4] Straka, M. et al. JMRI 2010;32:1024. **Acknowledgements:** This work was supported in part by the NIH (5R01EB002711, 5R01EB008706, 3R01EB008706, 5R01EB006526, 5R21EB006860, 2P41RR009784), the Center of Advanced MR Technology at Stanford (P41RR09784), Lucas Foundation, Oak Foundation, and the Swedish Research Council (K2007-53P-20322-01-4). An extra special thanks to Patricia Lassus, Murat Aksoy, and Nancy Fischbein for their insights.