Experimental considerations for OGSE of anisotropic tissue

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TARGET AUDIENCE: Experimentalists interested in oscillating gradient spin echo (OGSE) for mapping tissue microstructure with ultra short diffusion times.

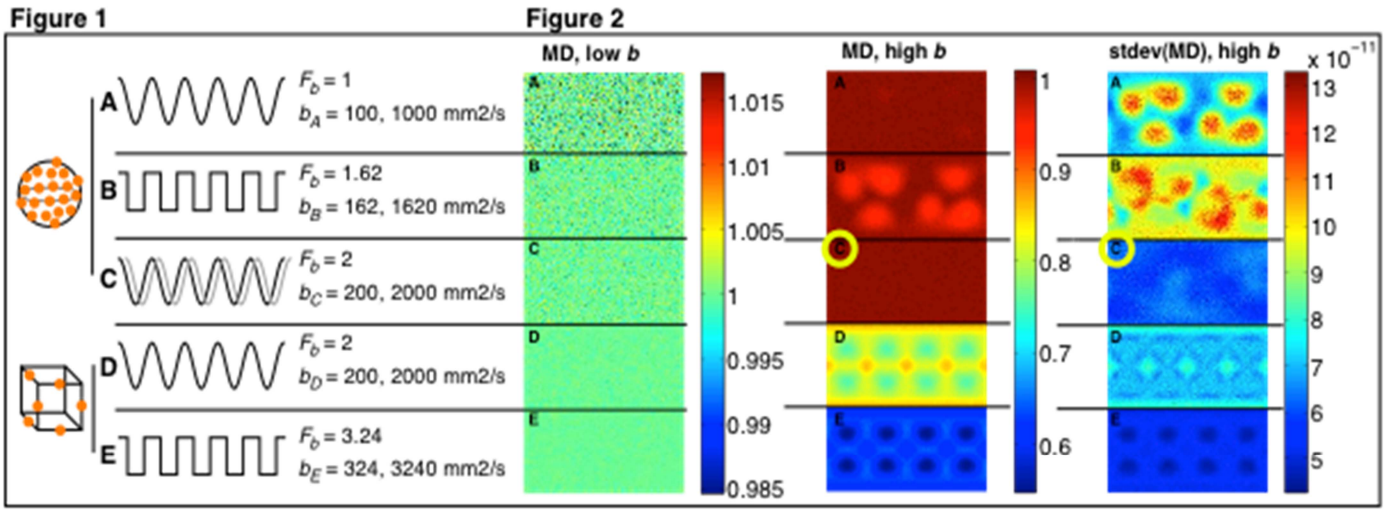
PURPOSE: The purpose of this study is to investigate different approaches for rotationally invariant sampling of oscillating gradient spin-echo (OGSE). It is yet to be investigated if a high number of directions or maximum diffusion weighting should be prioritized.

THEORY: Diffusion weighted imaging (DWI) with effective diffusion times below ~5 ms gives new contrasts to tissue microstructure and can effectively be explored with cosine modulated OGSE experiments[1]. The technique has recently been combined with diffusion tensor imaging (DTI) to extract rotationally invariant information from anisotropic tissues[2,3] and the first experiments on the human brain in vivo have been performed[4]. However, OGSE is complicated on conventional MRI scanners due to the low diffusion weighting *b*-value given by limited gradient strengths.

Different approaches can be used to increase the b-value of the conventional harmonic cosine modulated OGSE (A in figure 1) with minimal compromises on diffusion time specificity. Square/trapezoidal gradients give a *b*-value increase up to a factor of $F_b \approx 1.62$ depending on slew rate (B in figure 1)[4]. Planar diffusion encoding with circularly polarized OGSE (CP-OGSE) gives an increase by a factor of $F_b = 2$ [5]. These schemes (A-C) can be applied in any arbitrary orientation on a sphere limited by the maximum gradient amplitude in one direction as in HARDI acquisitions. By combining the maximum gradient strength in two directions simultaneously the *b*-value of the harmonic and square waves increase by $F_b = 2$ and $F_b = 3.24$ respectively. However this approach only allows 6 uniformly distributed directions and is fixed to the physical gradient system of the scanner. This is enough for fitting the six independent elements of the diffusion tensor but may induce a rotational bias as previously shown in conventional DTI analysis[6]. In this study we investigate the performance of the five OGSE schemes in figure 1A-E for a Gaussian diffusion tensor.

METHODS: Diffusion data was simulated analytically assuming an underlying Gaussian diffusion tensor as in Jones 2004 but with a Rician noise model as in magnitude MRI data[7]. A diffusion tensor **D** with mean diffusivity MD = $0.7\text{x}10^{-9}$ mm^2/s, FA = 0.8 and eigenvalues L1 > L2 = L3 was used for all simulations. The principal direction of **D** was rotated over the whole sphere relative to the gradient co-ordinate system. *b*-values were set to b_A= 100 and 1000 s/mm^2 for scheme A in figure 1. This is comparable to *b* – values achievable at frequencies around 100 Hz on clinical and preclinical MRI systems respectively[3,4]The corresponding diffusion weighting for the schemes B-E given the same gradient length and maximum gradient amplitude are scaled by the factor F_b and shown in figure 1. A and B were measured with 24 uniformly distributed directions and with one repetition, C was measured in planes orthogonal to the same 24 directions and with one repetition and D and E were acquired in 6 directions **g** = [x,y; -x,y; x,z; -x,z; y,z;-y,z], as seen in Figure 1, but repeated 4 times for comparable scan time and noise figures. Signal to noise was set to SNR = 25.

RESULTS: Figure 2 shows cylindrical projections (x/y-axes are azimuth/polar-angles) of the orientational variance in the MD estimates with the low and high *b*-values (left and middle) normalized to the true value, i.e. 1 equals accurate estimates. The standard deviation of the MD estimates from the high *b*-value is shown to the right, i.e. lower value yields higher precision. Similar patterns were observed for FA. For the low *b*-values, where b·MD << 1**,** the square wave scheme (E) with the highest effective *b* outperforms the schemes with higher angular resolution. At higher *b*-values, 6 gradient directions create a strong rotational variance (D and E) and higher angular resolution is needed to provide rotationally invariant estimates. The circularly polarized scheme (C, highlighted with yellow circle in the middle and right panel in figure 2) provides the most accurate and precise estimates.



DISCUSSION AND CONCLUSION: Our results suggest that clinical implementations of DTI with OGSE on conventional MRI systems with weak gradients should use square waves with gradients combined in six directions for maximizing diffusion weighting in few directions. With stronger gradient systems, as on preclinical or new experimental human scanners, CP-OGSE gives the best overall results both in terms of precision and accuracy. We emphasize that no universal solution guide the choice between the different solutions. For instance, the diffusion tensor may vary with diffusion time. Experimental design must be considered in relation to physiological and hardware limitations, SNR, the underlying tissue and the parameters of interest. Our approach provides a framework for such considerations.

REFERENCES: 1. Gore et al, 2010, NMR in Biomed, 2.Aggarwal et al, 2012, MRM, 3. Kershaw et al 2013, NeuroImage, 4. Van et al, 2013, MRM, 5. Lundell et al, 2013, in proc 21st ISMRM, 6. Jones 2004, MRM, 7 Guldbjartsson and Patz, MRM, 1995