3496

Impact of Residual Gradient Moments on Diffusion Weighted Imaging

Matthew J. Middione[1], Michael Loecher[1], Kévin Moulin[1], and Daniel B. Ennis[1]

[1]Radiological Sciences Lab, Department of Radiology, Stanford University, Stanford, CA, United States

Synopsis

Bulk motion corrupts diffusion measurements in the heart but can be mitigated by nulling first and second gradient moments. When using optimization methods to design diffusion gradient waveforms with moment nulling, imperfect gradient waveforms arise from discrete convergence criteria, which impart residual (non-zero) gradient moments. This leads to intravoxel dephasing, signal loss, and inaccurate ADC measurements. Herein, simulations show that residual $M_0 \leq 10^{-2}$mT/m·s, $M_1 \leq 10^{-4}$mT/m·s^2, and $M_2 \leq 10^{-5}$mT/m·s^3 leads to ≤5% increase in ADC. This work defines convergence criteria requirements for residual gradient moments that enable faster optimizations and more accurate measurements of ADC when using optimization methods for cardiac DWI sequence design.

Introduction

Diffusion-weighted imaging (DWI) measurements provide clinical value, but sensitivity to bulk motion frequently contributes to signal losses that can confound diffusion measurements, especially in moving tissues like the heart. Bulk motion sensitivity can be mitigated by using velocity (M_1=0) and/or acceleration-compensated (M_1=M_2=0) gradient waveforms to null the intravoxel phase dispersion from the bulk (coherent) motion of spins while preserving the intravoxel phase dispersion from incoherent diffusing spins. When using optimization methods to design diffusion gradient waveforms with moment nulling, imperfect gradient waveforms arise from discrete convergence criteria, which impart residual (non-zero) gradient moments. This leads to intravoxel dephasing, signal loss, and inaccurate ADC measurements. The intrinsic symmetry of conventional diffusion encoding strategies ensure exactly zero residual gradient moments as the waveforms are equal and opposite on either side of the refocusing pulse. However, optimization methods, which generate asymmetric gradient waveforms[1-5], are increasingly used for time optimal gradient waveform design and/or to add additional constraints that mitigate bulk motion, eddy currents, and concomitant field terms. In designing an optimization scheme, it is important to identify optimization criteria that enable fast solutions to be obtained. If convergence criteria are unnecessarily strict, then convergence times can be unacceptably long (i.e. minutes), whereas real-time (i.e. less than 10-50ms) solve times are preferred. The purpose of this work was to define acceptable limits for residual gradient moments that confer ≤5% measurement bias for ADC in the presence of bulk motion.

Methods

Numerical simulations were performed in Matlab to analyze the impact of residual gradient moments on intravoxel phase dispersion. The measured ADC was calculated for each of the 10,000 simulated spins per voxel as $ADC' = \frac{1}{-b} ln\left[abs\left(\sum_{n=1}^{10,000} \frac{S_{DWI}^n}{S_0}\right)\right]$, for b=1000mm^2/s, ADC=3x10^{-3}mm^2/s, and S_0=10,000, representing the cumulative signal within the voxel in the absence of diffusion encoding gradients and intravoxel phase dispersion. The diffusion weighted signal for the nth spin was calculated as $S_{DWI}^n = e^{-bADC} e^{-i\phi_n}$, with $\phi_n = \gamma r_n M_0 + \gamma v_n M_1 + \gamma a_n M_2$, γ as the gyromagnetic ratio, r_n, v_n, and a_n as the simulated position, velocity and acceleration of the nth spin, and M_0, M_1 and M_2 as the zero, first, and second order residual gradient moments. To analyze the impact of residual M_0, ADC' was computed for different pixel sizes (1-10mm) with residual values of M_0 and M_1=M_2=0. To analyze the impact of residual M_1, ADC' was computed for simulated intravoxel velocity gradients, of varying maximum velocities (0-0.2m/s), with residual values of M_1 and M_0=M_2=0. To analyze the impact of residual M_2, ADC' was computed for simulated intravoxel acceleration gradients, of varying maximum accelerations (0-1m/s^2), with residual values of M_2 and M_0=M_1=0. The following residual gradient moments were used in the simulations: 0, 10^{-6}, 10^{-5}, 10^{-4}, 10^{-3}, 10^{-2}, 10^{-1}, 10^{0}, and 10 with units of mT/m•s for M_0, mT/m•s^2 for M_1 and mT/m•s^3 for M_2.

Results

The impact of residual gradient moments on the measured ADC is shown in Figure 1 as a function of residual M_0 and pixel size (Figure 1A), residual M_1 and intravoxel velocity gradients (Figure 1B) and residual M_2 and intravoxel acceleration gradients (Figure 1C).

Discussion

Achieving precise gradient moment nulling is difficult. The simulations in this work show that knowledge of the pixel size and expected tissue motion can help define acceptable residual moment nulling thresholds while maintaining ADC measurements within 5%. The degree of intravoxel signal dephasing depends on the product of intravoxel velocity and acceleration gradients and any residual gradient moment(s). Residual M_0 will always lead to an increase in the measured ADC due to intravoxel signal dephasing. A negligible (≤5%) increase in the measured ADC is observed when the residual M_0 is on the order of 10^{-2}mT/m•s for all simulated pixel sizes (0-10mm) and 10^{-1}mT/m•s for simulated pixel sizes ≤4.5mm. Hence, higher resolution imaging is less prone to bulk motion artifacts. Residual M_1 and M_2 similarly lead to an increase in the measured ADC. A negligible (≤5%) increase in the measured ADC is observed when the residual M_1 is on the order of 10^{-4}mT/m•s^2 or when the residual M_2 is on the order of 10^{-5}mT/m•s^3. Defining acceptable thresholds for residual gradient moments is important when using convex optimized diffusion encoding gradient design approaches, where the residual moments need to be specified as a constraint in the optimization. Setting these values to a specific non-zero, but nearly zero value ensures minimal signal dephasing while also providing shorter optimization convergence times.

Conclusion

Residual gradient moments can lead to an increase in the measured ADC in DWI due to intravoxel signal dephasing. This work helps to define acceptable thresholds for residual gradient moments, which can be used to enable fast and more accurate measurements of ADC when using optimization methods for the pulse sequence design of diffusion sequences.

Acknowledgements

Funding NIH R01 HL131975 and HL131823 to DBE.

References

1. Aliotta et al. Convex optimized diffusion encoding (CODE) gradient waveforms for minimum echo time and bulk motion compensated diffusion weighted MRI. Magn Reson Med 2017;77:717–729.
2. Aliotta et al. Eddy current–nulled Convex Optimized Diffusion Encoding (EN-CODE) for distortion-free diffusion tensor imaging with short echo times. Magn Reson Med 2018;79:663–672.
3. Yang et al. Eddy current nulled constrained optimization of isotropic diffusion encoding gradient waveforms. Magn Reson Med 2018;00:1–15.
4. Sjölund et al. Constrained optimization of gradient waveforms for generalized diffusion encoding. J Magn Reson 2015;261:157-168.
5. Loecher et al. Accelerating 4D-Flow Acquisitions by Reducing TE and TR with Optimized RF and Gradient Waveforms. ISMRM 2018

Figures

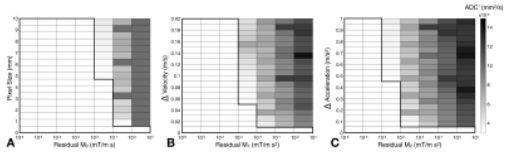

Figure 1: Numerical simulations showing the impact on the measured ADC (ADC') arising from intravoxel phase dispersion (signal loss) as a function of residual M_0 and pixel size (**A**), residual M_1 and intravoxel velocity gradients (**B**), and residual M_2 and intravoxel acceleration gradients (**C**). Measured ADC' values that vary ≤5% compared to the simulated ADC ($3x10^{-3}mm^2/s$) and b-value ($1000mm^2/s$), are highlighted by the black borders (lower-left area within plots).