Characterization of high performance human gradient system for spin echo cardiac DTI

Konrad Schieban[1], Timothy G Reese[2], Christian T Stoeck[1], David E Sosnovik[2], Sebastian Kozerke[1,3], and Choukri Mekkaoui[2]
[1]*Institute for Biomedical Engineering, ETH Zurich, Zurich, Switzerland,* [2]*Radiology, Harvard Medical School, Massachusetts General Hospital, Martinos Center for Biomedical Imaging, Charlestown, MA, United States,* [3]*Division of Imaging Sciences, King's College London, London, United Kingdom*

Introduction Cardiac DTI is complicated by the effects cardiac bulk motion on the diffusion tensor. To overcome the impact of motion, sequences encoding diffusion over two consecutive cardiac cycles using stimulated echoes have been developed (STEAM, [1]). Despite its robustness to motion, STEAM has the downside of an inherently low SNR and it has been shown that myocardial strain interacts with the measurement of the diffusion tensor [2]. The development of high-performing gradient coils with field strengths of over 80mT/m, raises the possibility of replacing the unipolar gradient pulses in the Stejskal-Tanner sequence [3] with bipolar gradient pulses which compensate for the first-order Taylor-approximation of motion [4]. The advantage of this approach is that a spin echo, rather than STE, readout can be used. However, the benefits of powerful gradient coils might be offset by the effect of eddy currents, which become increasingly influential with large gradient amplitudes. Whole body human scanners with gradient strengths up to 300mT/m have recently been developed. The aim of this work was therefore to determine whether increasing the gradient amplitude from 80mT/m to 150mT/m would greatly decrease the measurement error of the diffusion tensor and if the advantage of larger gradient amplitudes would be diminished by the increase in eddy currents.

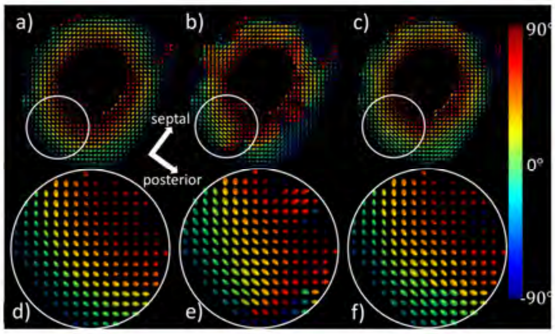

Figure 1 Impact of motion for different gradient field strengths. The color of the glyphs corresponds to the helix angle. a,d) Undistorted reference. The impact of motion is shown for G=80mT/m (b,e) and G=150mT/m (c,f).

Methods Left-ventricular motion was analyzed using 3D-tissue tagging together with the 3D SINMOD approach [5] to estimate displacement fields for tissue points in the myocardium. In the presence of diffusion-sensitizing gradient pulses, moving sets of nuclear spins accrue phase, resulting in signal attenuation due to spin incoherence inside voxels. The associated measurement error of the DTI-indices MD, FA and the helix angle [6] was evaluated in computer simulations for a voxel resolution of $2.7 \times 2.7 \times 8mm^3$. Eddy current measurements were performed on a commercial 3T scanner with a modified 300mT.M whole-body gradient (MAGNETOM Skyra CONNECTOM, Siemens Healthcare, Erlangen, Germany). The spatiotemporal dependency of eddy current-induced fields of a bipolar gradient pulse with $b = 500s/mm^2$, $G = 150mT/m$ and a rise time of $T_{Rise} = 16\mu s/(mT/m)$ was measured using PRESS [7] and modeled with a set of spherical harmonic functions up to 2nd order [8]. The effect of eddy currents in cardiac DTI was demonstrated by imposing the measured phase on the k-space matrix of six reference DWIs in numerical simulations.

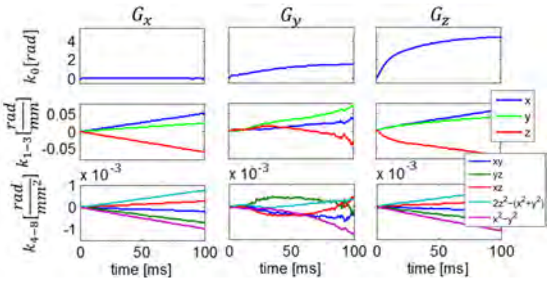

Figure 2 Spatial orders of eddy current-induced phase due to a bipolar pulse in the x, y and z-directions.

Experimental Geometric distortions due to eddy currents and their influence on the measured ADC were demonstrated in phantom measurements. The phantom was imaged with $TE = 74ms$, $TR = 3s$, a resolution of $2.5 \times 2.5 \times 5mm^3$, a matrix-size of 96×78 and a b-value of $500s/mm^2$. The signal was acquired using an EPI readout with an acceleration factor of 2 and reconstructed with GRAPPA [9].

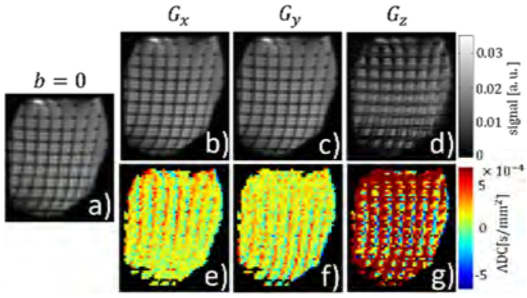

Figure 3 Reference phantom image (a) and DWIs in x-, y- and z-direction (b-d). Estimates of the ADC are shown in e)-g).

Results Fig.1 illustrates the distortion of the diffusion tensor in the myocardium due to motion for different gradient strengths. It can be seen that the helix angle distribution of the reference image (1a) is substantially distorted at 80mT/m (1b) but only minimally distorted at 150mT/m (1c). The spatiotemporal dependency of eddy current-induced phase is shown in Fig.2. It can be seen that the pulse in z-direction creates a strong B_0-offset with a mono-exponential decay constant of 18ms, inducing a phase of approximately π within the first 20ms. Diffusion weighted images of the phantom are shown in Fig.3. There is hardly any visual difference between the reference image (3a) and the DWIs in x- and y-direction (3b,c) but strong image distortion can be seen in the DWI in z-direction (3d). Measurements of the ADC in x-,y- and z-direction are shown in 3e)-g). The large difference in the ADC estimate in z-direction (3g) cannot be related to the phantom structure but to geometric distortions in the DWI. In Fig. 4, the error of MD, FA and the helix angle (HA) due to motion and eddy currents with B_0-correction are shown for direct comparison. It can be seen that linear and higher order eddy current field terms only create a small error in the DTI-indices. Global error averages over an axial mid-slice through the left ventricle are shown in Table 1.

	Motion G= 80mT/m	Motion, G= 150mT/m	Eddy Currents B_0-corrected
MD[%]	18.86	9.91	5.12
FA	0.075	0.041	0.052
HA[deg]	16.84	7.53	6.83

Table 1 Averaged error of DTI-indices in the LV wall.

Discussion In this work, we show that cardiac motion has a substantial impact on DTI measurements at gradient strengths of 80mT/m or less but the error of DTI-based indices can be substantially attenuated at 150mT/m. The error due to spin dephasing can also be mitigated by choosing a smaller voxel size but only by penalizing the SNR. An important factor that needs to be considered at 150mT/m are eddy currents. Geometric distortions due to a temporally nonlinear B_0-offset prevented an accurate estimate of the ADC in both phantom experiments and computer simulations. The reason might be the inductive coupling of the gradient coil in z-direction with the cryostat which generates a magnetic field in the same direction. Linear and higher order field terms, in contrast, were found to be benign. In the experiments we did not exploit the performance of the gradient coils which can create field strengths of up to 300mT/m and can also slew considerably faster. The reason is that the nerve stimulation threshold upper bounds the slew rate and hence, gradient pulses with peak amplitude of 300mT/m do not yield appropriate b-values for cardiac DTI. Even though PRESS is useful to characterize residual fields of lower spatial order, spoiler and localization gradients may create eddy currents as well and thus, interfere with the measurement. A more accurate technique which overcomes this measurement bias is the use of an NMR field camera [10]. Cardiac DTI at 150mT/m is highly promising to advance research, yet, it is necessary to account for the decaying B_0-offset due to eddy currents in image reconstruction.

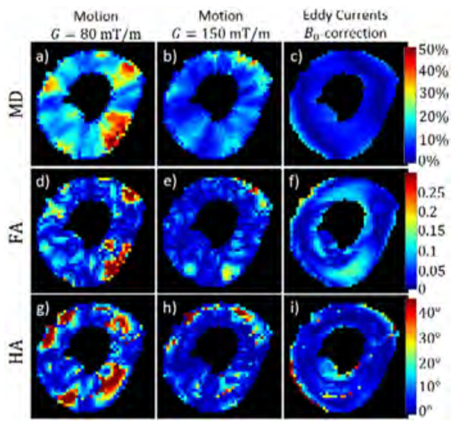

Figure 4 Error of MD, FA and HA caused by motion and eddy currents after B_0-correction.

References **[1]** Reese TG, MRM 34 (1995) 786. **[2]** Reese TG, JMR 112 (1996) 253. **[3]** Stejskal EO, JChemPhys. 42 (1965) 288. **[4]** Gamper U, MRM 57 (2007) 331. **[5]** Arts T, MedImag 29 (2010) 1114. **[6]** Scollan DF, AJP 275 (1998) 2308. **[7]** Bottomley PA, Ann NY Acad Sci 508 (1987) 333 **[8]** Barmet C, MRM 62 (2009) 269. **[9]** Griswold MA, MRM 47 (2002) 1202. **[10]** De Zanche N, MRM 60 (2008) 176.