Increasing the resolution of diffusion-weighted MRI with distortion compensated orthogonal acquisitions and super-resolution reconstruction

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Introduction. Increasing the spatial resolution in diffusion-weighted imaging (DWI) is very challenging with a single shot EPI acquisition. It requires sampling of higher frequencies in k-space, making the acquisition highly demanding on the scanner gradient coils when achieving the spatial encoding. Additionally, reducing the voxel size by a factor α is challenging because it requires α^2 averages to ensure a similar SNR. Ultimately, the time to encode a larger k-space is not negligible and leads to larger echo time (TE). First this leads to severely increased geometric and intensity distortion in the phase encoding direction due to T2* relaxation. Second, because the DW-signal exponentially decreases with increasing TE, this strongly impacts the SNR as well. Recently, acquisition of orthogonal *anisotropic* acquisitions and super-resolution reconstruction (SRR) of the underlying high-resolution acquisition has been proposed to achieve higher resolution by reducing the spatial encoding burden of each acquisition [1]. Promising results have been shown with simulations. However, evidence that SRR enables resolution enhancement remains unclear in practice. Particularly, orthogonal acquisitions are subject to very different distortion, making the precise alignment of the images impossible and strongly perturbing the reconstruction. In this work we investigate for the first time the SRR from distortion compensated real orthogonal DWI acquisitions. We demonstrate that combining distortion compensation and SRR provides better results than acquisition of a single isotropic scan for the same acquisition duration time. This provides the first evidence that SRR may enable resolution enhancement in DWI.

Material and methods. We acquired K=3 anisotropic DWI scans (axial, coronal, and sagittal) on a Siemens 3T Trio with a 32 channel head coil (FOV=220mm, matrix=176x176, in-plane resolution=1.25x1.25mm^2, slice-thickness=2.5mm, 5 b=0s/mm^2, 30 directions at b=1000s/mm^2, total scan time=17min). For comparison, we acquired an isotropic DWI scan with parameters chosen to match the acquisition time of the three previous scans (same parameters except matrix=146x146, resolution=1.5x1.5x1.5mm^3, two averages, 16min32). Finally, a dual echo gradient echo field map image was acquired (TE$_{1/2}$=5.19/7.65ms, resolution=2x2x2.5mm^3). Each acquisition was compensated for distortion via field map unwarping [2] by using the phase field map of the dual echo gradient scan. Possible patient motions were corrected by aligning the DW-volumes both in space and q-space, so that the DW images are aligned and correspond to the exact same gradient orientation set in the patient coordinate system. Following the approach of [1], the super-resolution reconstruction (SRR) was formulated as a MAP problem. It relies on a volume acquisition model which describes the generation of the observed scans from the unknown high-resolution image. It enables the introduction of image priors that exploit spatial homogeneity and enables regularized solutions. The estimation of the high-resolution images was performed via a gradient minimization descent. The SRR of each DW-image was achieved to create an isotropic volume with voxel size 1.25x1.25x1.25mm^3 (FMC-SRR). To investigate the effect of the distortion compensation, we also achieved the SRR without any field-map correction (SRR) at the same isotropic resolution. We compared the SRR technique to the isotropic acquisition by resampling the 1.5x1.5x1.5mm^3 DW-images to 1.25x1.25x1.25mm^3 (ISO). Finally, we investigated the effectiveness of applying a postprocessing denoise correction technique [3] to the isotropic acquisition (d-ISO). We qualitatively compared the color-FA maps of ISO, d-ISO, SRR and FMC-SRR, and compared the tractography results of the corpus callosum achieved from the same seeding region. We qualitatively compared the four experiments via residual bootstrapping in a common 3-D region of interest in the external capsule.
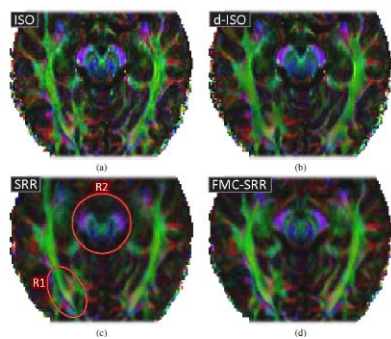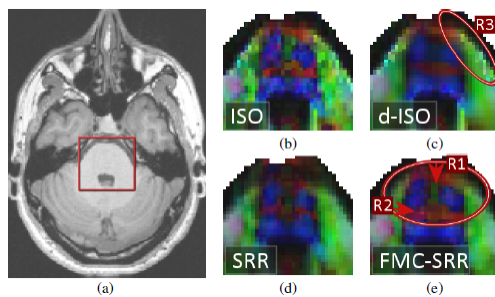

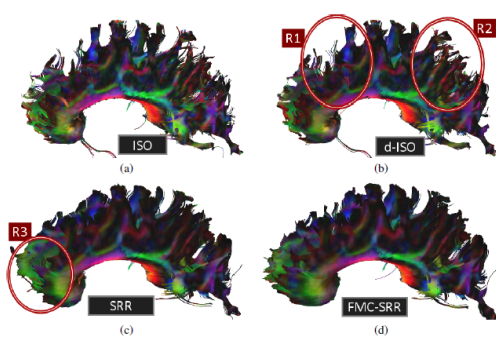Fig. 1. Color-FA.　　　Fig. 2. Zoom in a region of the brain stem.　　　Fig.3. Tractography results for the corpus callosum.

Results. Fig.1 reports the color-FA maps for ISO, d-ISO, SRR and FMC-SRR. ISO (Fig.1a) provides a noisy color-FA, which is corrected when applying the post-processing noise correction (d-ISO, Fig.1b). However, fines structures are smoothed and lost when applying the noise correction (see Fig.1b, R1). Fig.1c shows that non-compensation of the distortion (SRR) leads to a blurred color-FA map, particularly in the region R2. Fig.1d shows that fine structures far smaller than the slice thickness (2.5mm) are well conserved with FMC-SRR (region R1). Importantly, structures appear to be more detailed with FMC-SRR than with d-ISO (Fig.1d R1/R2). Fig.2 shows a zoom on the color FA in a region of the brain stem. Consistently with Fig.1a, ISO provides a highly noisy color-FA (Fig.2b). d-ISO provides more smoothed results (Fig.2c) but with interpolation artefacts (R3) and more blurred structures than FMC-SRR (region R1). Fig.2d (SRR) shows that misalignment of the acquisitions due to distortion leads to a truncated structure (R2). Fig.3 reports tractography results for the corpus callosum. ISO (Fig.3a) provides poor results due to the high noise corruption. The noise filtering in d-ISO (Fig.3b) enables a better connectivity assessment. However, a number of streamlines remains disorganized (Fig.3b R1/R2). FMC-SRR (Fig.3d) outperforms the other approaches. Particularly, streamlines in the frontal part (R3) better represent the anatomy. Without distortion correction (SRR, Fig.3c), the anisotropic scans were poorly aligned in the frontal region and the tracts prematurely stopped. Finally, the FA variance estimated by bootstrapping was 3.9e-2 for ISO and respectively 2.08e-2, 2.09e-2 and 2.07e-2 for d-ISO, SRR and FMC-SRR.

Conclusion. Instead of employing a dense high-resolution sampling of k-space, we reduce the spatial encoding burden by employing multiple scans with high-resolution only along a limited number of axes. Because acquisitions with different phase-encoding directions experience very different distortion due to T2* relaxation, we compensate for distortion to ensure that overlapping voxels across acquisitions represent the same brain location. We reconstruct the underlying high-resolution acquisition via super-resolution reconstruction. We demonstrate that our FMC-SRR approach provides better results than acquisition of a single isotropic scan for the same acquisition duration time. Denoising the isotropic acquisition provides comparable estimation uncertainty to the SRR technique but, as seen in Fig.1 and Fig.2, leads to smoothed and lost fine structures. The SRR provides more detailed structures and better tractography results. Performing high-resolution sampling of k-space along only a limited number of axes enables imaging with higher SNR and lower distortion. More importantly, this enables to go beyond the minimum isotropic resolution achievable by the scanner, and may lead to DWI imaging with unprecedented isotropic resolution. This work provides the first evidences that SRR, which employs *conventional* single-shot EPI techniques, may enable resolution enhancement in DWI, and may dramatically impact the way to achieve DW imaging in both neuroscience and clinical applications.

References.

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