Impact of the point-spread function on parameters derived from diffusion-weighted imaging: axial versus sagittal acquisition

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Introduction: Diffusion-Weighted (DW) MRI is fast becoming the method of choice for investigations of brain white matter and its connectivity. Due to the intrinsic sensitivity of the phase of the signal to random microscopic motion, single-shot imaging techniques are typically used, with echo-planar imaging (EPI) being by far the most common. DW data intended for tractography or voxel-based analyses, which are inherently 3D, are commonly acquired with isotropic voxels (i.e. slice thickness = in-plane voxel size) and contiguous slices. While this approach minimises any bias due to differences in voxel size along the three axes, there are remaining differences in the point spread function (PSF) between these directions. While unavoidable, these differences can introduce differences that depend on the orientation of brain structures with respect to the imaging plane. In this study, we demonstrate the impact of these differences in the PSF by comparing DW images acquired in the sagittal plane with the equivalent images acquired in the axial plane.

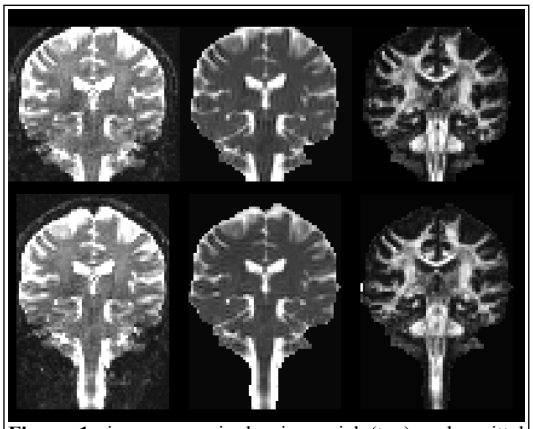

Figure 1: images acquired using axial (top) and sagittal (bottom) acquisitions, displayed in the coronal plane. From left to right: *b*=0 image, ADC map, and FA map.

Methods: Data were acquired from a healthy volunteer on a 3T Siemens Trio system using a DW twice-refocused echo-planar imaging (EPI) sequence, with the following parameters: FOV=216×216mm, matrix size=90×90, 56 contiguous slices, 2.4mm slice thickness, voxel size=2.4×2.4×2.4mm, 25 DW directions + 5 b=0 volumes, b=1000 s/mm^2, parallel imaging acceleration factor 2, and 2× oversampling along the RO direction. For the axial data sets, the RO direction was left-right, and the phase-encode (PE) direction was anterior-posterior. For the sagittal data sets, the RO direction was inferior-superior, and the PE direction was also anterior-posterior. 7 repeats of each data set were acquired, with the acquisition alternating between axial and sagittal to minimise differences due to scan-to-scan movement. For each acquisition protocol, the positions of the acquired volumes of interest were aligned to ensure voxel alignment between the axial and sagittal acquisitions.

The data were processed using the diffusion tensor model to produce mean diffusivity (MD) and fractional anisotropy (FA) maps [1]. In addition to visual inspection, significant differences between axial and sagittal acquisitions were investigated by extracting the region of overlap between the acquired volumes, and performing voxel-wise independent t-tests (assuming equal sample sizes and equal variance, and applying a Bonferroni correction for all voxels within a brain mask). The statistical analysis was performed for all 7×5 *b*=0 images, and the 7 generated FA and ADC maps.

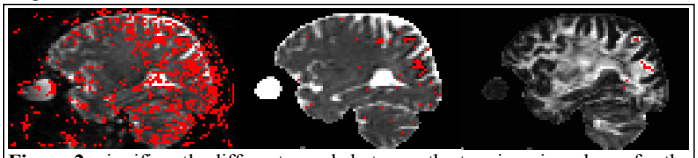

Figure 2: significantly different voxels between the two imaging planes for the *b*=0 images (left), ADC maps (middle) and FA maps (right). Significant voxels are shown in red, overlaid on the corresponding axial image.

Results: Images from the two imaging planes are visually indistinguishable in terms of SNR or susceptibility-induced distortions (Figure 1). However, statistical analysis of *b*=0 images reveals significant differences in a large number of voxels (Figure 2). Significant differences were also found for the ADC and FA maps, although the number of affected voxels is much smaller (especially for the FA). Close inspection of the *b*=0 images reveals that most of the observed differences coincide with sharp edges, particularly CSF/brain interfaces (Figure 3). These differences are likely due to differences in the point spread function (PSF) along the slice-select (SS) direction compared to the RO direction, giving rise to different patterns of Gibbs ringing. The voxels found to be significantly different in the ADC and FA analyses were also predominantly located in such regions.

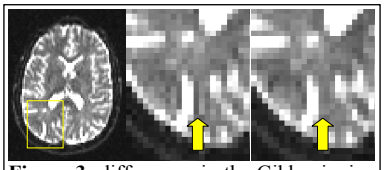

Figure 3: differences in the Gibbs ringing artefact in the *b*=0 image, for the region corresponding to the significant FA cluster in Figure 2. Left: axial *b*=0 image showing the location of the region of interest. Middle: the magnified *b*=0 image acquired using the axial acquisition. Right: the magnified *b*=0 image acquired using the sagittal acquisition. Note the Gibbs ringing artefact in the axial acquisition as highlighted by the arrow.

Discussion: The sagittal acquisition protocol used in this study was designed to be equivalent in all aspects to the axial protocol. The PE direction was along the same direction (anterior-posterior in this case), leading to identical susceptibility-induced distortions. Susceptibility-induced signal drop-out is negligible due to the use of a spin-echo sequence (in contrast to the gradient-echo case, where signal drop-out is dependent on the slice orientation). The availability of oversampling along the RO direction on modern systems allows for an increased FOV along this direction with no other modifications to the sequence (in fact, oversampling along the RO is performed by default on many current scanners); wrap-around along the RO direction (inferior-superior in this case) is therefore not an issue. Finally, the RO axis was along the inferior-superior direction, meaning that the Z gradient was most rapidly switching (as opposed to the X for the axial case). Since the Z gradient has been shown to have the highest threshold for peripheral nerve stimulation [2], the EPI gradients can be switched at least as rapidly as in the axial case, leading to similar bandwidth per pixel.

There were however significant differences between the two acquisitions, mostly in voxels adjoining sharp edges in the signal. These differences can therefore be attributed mostly to differences in the PSF along the RO compared to the SS direction. These were most pronounced for the *b*=0 images, due to the large signal difference between CSF and tissue (Figure 3). Although these differences do exist, this by no means implies that one imaging plane is better than the other, since ringing artefacts will be present in both, but along different directions. It does however show that the signal intensity in DW images and derived parameter maps is significantly affected by the PSF, and that sharp edges (particularly in the *b*=0 image) will affect adjoining voxels differently along the SS and RO directions.

The results suggest that ADC is more sensitive to Gibbs ringing than FA. The reason for this is likely due to the different effect of Gibbs ringing on the *b*=0 signal compared to the DW signal in voxels adjoining CSF. In the *b*=0 image, CSF is much brighter than brain, and the ringing will tend to reduce the *b*=0 signal in adjoining voxels. Since the ADC is related to the ratio of the DW signal to the *b*=0 signal, this will lead to significant errors in these regions. On the other hand, the FA is related to differences in the ADC along different directions, normalised to the root-mean-square ADC. The differences introduced by the Gibbs ringing will therefore tend to cancel out, with any remaining differences likely related to the non-linear relationship between signal and ADC.

Besides changing the direction of the PSF, using a sagittal acquisition may have other advantages. First, it results in increased coverage, allowing a significant portion of the cervical spine to be imaged with no increase in scan time or degradation in image quality. Second, it may be advantageous in terms of tolerance to motion: the most common forms of motion are a rotation in the sagittal plane, or a translation in the inferior-superior direction. These tend to have serious consequences for axial acquisitions due to spin history effects. Images acquired sagittally may be less prone to such artefacts, since these types of motion would then remain in-plane.

Conclusion: Differences in the PSF along the slice direction versus in-plane directions have a significant effect on DW images and derived parameters. This introduces differences in the measured parameters that depend not only on the presence of sudden signal intensity differences (particularly tissue/CSF interfaces), but also whether these structures are within-plane or in an adjacent slice. Although these artefacts cannot be avoided, they can be minimised in a location of interest by tailoring the imaging plane to ensure the direction of greatest signal change is through-slice (see e.g. figure 3). In addition, it should be noted that far from being sub-optimal, the use of a sagittal acquisition plane may have certain advantages over the axial equivalent without compromising image quality.

References: [1] Basser PJ, NMR Biomed 8: 333-44 (1995). [2] Faber SC, et al., MRI 21: 715-24 (2003). [3] Alexander DC, et al., NeuroImage 52: 1374-89 (2010).