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k-space and q-space: Combining ultra-high spatial and angular resolution in diffusion imaging using ZOOPPA at 7 T

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ABSTRACT

There is ongoing debate whether using a higher spatial resolution (sampling k-space) or a higher angular resolution (sampling q-space angles) is the better way to improve diffusion MRI (dMRI) based tractography results in living humans. In both cases, the limiting factor is the signal-to-noise ratio (SNR), due to the restricted acquisition time. One possible way to increase the spatial resolution without sacrificing either SNR or angular resolution is to move to a higher magnetic field strength. Nevertheless, dMRI has not been the preferred application for ultra-high field strength (7 T). This is because single-shot echo-planar imaging (EPI) has been the method of choice for human in vivo dMRI. EPI faces several challenges related to the use of a high resolution at high field strength, for example, distortions and image blurring. These problems can easily compromise the expected SNR gain with field strength. In the current study, we introduce an adapted EPI sequence in conjunction with a combination of ZOOmed imaging and Partially Parallel Acquisition (ZOOPPA). We demonstrate that the method can produce high quality diffusion-weighted images with high spatial and angular resolution at 7 T. We provide examples of in vivo human dMRI with isotropic resolutions of 1 mm and 800 μm. These data sets are particularly suitable for resolving complex and subtle fiber architectures, including fiber crossings in the white matter, anisotropy in the cortex and fibers entering the cortex.

Introduction

Diffusion magnetic resonance imaging (dMRI) is currently the most important tool for investigating white matter structures within the living human brain; see for example (Jones, 2008). It has been shown that at a voxel resolution of around 2–3 mm, a simple tensor model can be used to track the major white matter pathways in the human brain (Mori and van Zijl, 2002). However, this way it is impossible to accurately track through regions with more complex fiber arrangements such as crossing, fanning and branching. One way to address this issue would be to increase the spatial resolution (extending the acquisition in k-space), thereby reducing the number of concurrent fiber populations per voxel. However, increasing the resolution also means decreasing the signal-to-noise ratio (SNR), and thus impairing the robustness of fiber tract reconstruction. Therefore, high spatial resolution has so far been achieved at the expense of; (1) applicability to living humans (McNab et al., 2009; Miller et al., 2011; Roebroeck et al., 2008), (2) isotropy, by using thick slices (Finsterbusch, 2009; Karampinos et al., 2009; Yassa et al., 2010), or (3) angular resolution (Sarlls and Pierpaoli, 2009). Alternatively, one can apply more sophisticated local models, which are able to model multiple fiber populations and rely on higher angular resolution and/or multiple b-values (extending the acquisition in q-space) (e.g. Behrens et al., 2003; Jansons and Alexander, 2003; Tournier et al., 2004; Tuch, 2004; Tuch et al., 2002; Wedeen et al., 2005). For example, in diffusion spectrum imaging (DSI), a low spatial resolution of about 3 mm is used to achieve sufficient SNR to enable the acquisition of a high number of diffusion directions and multiple b-values, thereby resolving crossing fiber directions.

Since the MR signal scales with the main magnetic field strength, one possible way to increase the spatial resolution without sacrificing SNR or angular resolution is to move to a higher magnetic field strength. In theory, a gain of more than a factor of two in SNR can be expected when moving from a 3 Tesla to a 7 Tesla MR scanner. Even though applications such as high resolution anatomical imaging and functional MRI are already benefitting from the higher field strength, dMRI has not been seen as a preferred application for ultra-high field imaging yet. This is due to inherent problems associated with high resolution diffusion-weighted single-shot EPI (ssh-EPI) at high field strength. The two major challenges, which scale with the field strength and the resolution, are susceptibility induced distortions and image blurring caused by T_2^* relaxation. These effects can easily compromise the expected SNR gain for dMRI. Even though these negative aspects have been identified and are well understood,

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ssh-EPI is still the most frequently used methodology for human in vivo dMRI.

The above-described challenges can be addressed by using parallel imaging methods, as was first shown by Griswold et al. (1999). In parallel imaging, also known as Partially Parallel Acquisition (PPA), the MR data is acquired in parallel by multiple independent receiver coils. This allows the acquisition to be accelerated by skipping a certain fraction of MR data, followed by a reconstruction procedure. Typically, the acceleration factor (AF) ranges between two and three, meaning that the echo train length is reduced accordingly. In the reconstructed data set, after the parallel image reconstruction, the time between consecutive echoes, the acquired and the reconstructed echo, is the effective echo spacing. This is the echo spacing of the acquired data divided by the AF. Hence, because EPI distortions linearly depend on the time between consecutive echoes, they are reduced by parallel imaging. Additionally, the shortened EPI readout duration reduces image blurring, allowing a higher spatial resolution (Griswold et al., 1999) and improves the SNR. However, this gain in SNR is reduced by an intrinsic SNR loss due to the reduced data sampling, amounting to the square root of the AF. Furthermore, an additional SNR loss related to the parallel image reconstruction procedure using receiver coils with sub-optimal coil geometries has to be taken into account. This additional SNR loss is described by the g-factor (Pruessmann et al., 1999), which is always greater than or equal to one. More specifically, the SNR is decreased by $1/(g\text{-}factor\times\sqrt{AF})$. Please note that the g-factor is in turn a non-linear function of the AF. For higher acceleration factors, in particular, the g-factor increases drastically.

More than a decade before parallel imaging was used to accelerate in vivo MRI, inner volume imaging (Feinberg et al., 1985) was introduced to speed up the acquisition by acquiring a reduced field-of-view (FOV). A reduced FOV corresponds to a reduced sampling density in k-space. In other words, a certain fraction of time-consuming phase-encoding steps can be skipped, resulting in accelerated acquisition. When the reduced FOV is smaller than the object to be imaged, wraparound artifacts, also known as aliasing artifacts, will affect the resulting image. In inner volume imaging, aliasing artifacts are avoided by exciting and refocusing only the reduced FOV covering the region of interest.

The inner volume approach was applied to EPI to achieve a higher spatial resolution instead of reducing the acquisition time (Feinberg and Hale, 1986), later called zonal EPI and zoomed imaging (Mansfield et al., 1988). This was achieved by keeping the number of phase encoding steps constant. Similar to parallel imaging, reduced FOV imaging is affected by an intrinsic SNR loss due to reduced data sampling, which is related to the AF. However, reduced FOV imaging does not suffer from the additional losses due to the g-factor penalty. The major disadvantage of reduced FOV imaging is the limited imaging region. When large AFs are required, a very small FOV has to be chosen.

In the current work, we use a combination of a reduced FOV acquisition (ZOOmed imaging) and Partially Parallel Acquisition (PPA), named ZOOPPA, to benefit from the advantages of both methods and to improve the image quality of single-shot EPI (Heidemann et al., 2008). For this purpose, we implemented ZOOPPA in a diffusion-weighted EPI sequence (Heidemann et al., 2009), which is used to perform isotropic high resolution diffusion MRI (dMRI) at ultra-high field strength (7 T). In conjunction with a high performance gradient system, this approach enables dMRI with an isotropic resolution down to 800 μm. The dMRI data acquired have sufficient SNR to resolve complex fiber configurations in white matter. In addition, the diffusion signal in the cortex shows clear radial anisotropy and the high spatial resolution allows the detection of subcomponents of white matter fiber bundles, such as fibers entering the cortex, which are difficult to observe with MRI.

Theory and methods

Accelerated MRI

As mentioned in the Introduction, reduced FOV imaging can be used to accelerate the MR data acquisition. Parallel imaging is also used to speed up the acquisition. With regard to the k-space trajectory or sampling density, reduced FOV imaging and parallel imaging are identical. In both cases, a reduced FOV is acquired by reducing the k-space sampling density. The major difference lies in the way aliasing artifacts, resulting from undersampling, are avoided. In parallel imaging, after data acquisition, a reconstruction procedure is applied to the undersampled data in order to obtain a final image without aliasing. This is done to obtain a full FOV image. In comparison, for reduced FOV imaging, a pre-experiment is used to ensure that no signal is received from the region outside the reduced FOV before the data acquisition. For this pre-experiment, there are two main options. Besides the inner-volume imaging methods including localized excitation (Feinberg et al., 1985; Mansfield et al., 1988; Pauly et al., 1989), outer-volume suppression (OVS) (Le Roux et al., 1998) can be used. For outer-volume suppression, tissue located outside the reduced FOV is excited and subsequently suppressed. The OVS method is less sensitive to B_1 inhomogeneities than inner-volume excitation, a feature that is especially advantageous for ultra-high field imaging (Pfeuffer et al., 2002). As in parallel imaging, the reduced FOV approach allows a reduction of the EPI read-out time, resulting in reduced distortions and T_2^* blurring. However, reduced FOV imaging can be affected by remaining signal from the region outside the reduced FOV, which will result in aliasing artifacts. Furthermore, the low coverage, especially with high AF, is a disadvantage.

In fMRI, a high in-plane resolution has been achieved with both inner-volume imaging (Duong et al., 2002) and outer volume suppression (Pfeuffer et al., 2002). For diffusion MRI, inner-volume imaging has been used for ADC mapping of the human optic nerve (Wheeler-Kingshott et al., 2002) and later, OVS was used for spinal cord dMRI (Wilm et al., 2007). In both studies, the FOV was very small, 30 mm and below, a low number of slices were acquired and highly anisotropic voxel sizes were used. This was acceptable because the objects of interest, the optic nerve and the spinal cord, fit within this small FOV and the structure does not change abruptly along the main axis of the object. In this case, thick slices will not be significantly affected by partial volume effects and a low number of slices are sufficient to cover the main structure.

ZOOmed Partially Parallel Acquisition (ZOOPPA)

To overcome the weakness of each separate technique, reduced FOV imaging can be combined with partially parallel acquisition methods. This technique, given the name ZOOPPA (ZOOmed Partially Parallel Acquisition), was originally developed and implemented for fMRI (Heidemann et al., 2008) with gradient-echo EPI. We have adapted ZOOPPA for diffusion-weighted spin-echo EPI (Heidemann et al., 2009). In the current study, we use ZOOPPA to achieve high acceleration factors which enable the acquisition of high resolution single-shot EPI data with a relatively large FOV. Here, part of the overall acceleration comes from the zoomed approach, e.g. by acquiring 50% FOV with OVS (see Fig. 1B) resulting in an acceleration factor due to zoomed imaging $AF_{ZOOM}=2$. In the next step, this reduced FOV is further reduced by parallel imaging, e.g. acquiring only one third of the reduced FOV (see Fig. 1C) resulting in an acceleration factor due to parallel imaging $AF_{PPA}=3$. In total, only a sixth of the FOV is acquired. In this example, the overall acceleration factor would be $AF=AF_{ZOOM}*AF_{PPA}=2\times3=6$. Due to the parallel image reconstruction applied, 50% of the FOV will be obtained (as in Fig. 1B) with $AF=6$, instead of only 20% of the FOV when only the zoomed approach is used.