MR Imaging of ^{n}X-Nuclei (^{23}Na & Friends):
From Controversies to Potential Clinical Applications
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Image Reconstruction Issues Related to Real Resolution, Partial Volume & B_1/B_0 Corrections

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Highlights/ take-home messages

- The effective spatial resolution is influenced by the k-space trajectory, filtering, and fast transverse relaxation
- Correction of B_1-inhomogeneities can improve quantitative accuracy of X-nuclei MRI
- Quantitative imaging of the sodium concentration often requires correction of partial volume effects
- Iterative image reconstruction can markedly improve overall image quality

Introduction

Quadrupolar X-nuclei such as ^{23}Na, ^{17}O, ^{35}Cl, and ^{39}K experience a strong interaction with local electric field gradients. For spin 3/2-nuclei, this results in rapid biexponential signal decay. Transverse relaxation times of ^{23}Na are in the order of a few milliseconds (short/long component $T_{2,s}^{*}$ / $T_{2,l}$* ≈ 0.2-5.0 ms/ 10-64 ms) (1). Transverse relaxation times of oxygen (^{17}O), chlorine (^{35}Cl) and potassium (^{39}K) can be even shorter (2). Thus, acquisition techniques that enable ultra-short echo times (UTE) (3) are a prerequisite for quantitative X-nuclei MRI. For instance, conventional three-dimensional radial sampling of k-space can be employed (4-6). However, higher signal-to-noise ratio (SNR) can be achieved with advanced UTE techniques such as density-adapted projection reconstruction (DA-3D-PR) (7,8), twisted projection imaging (TPI) (9,10), acquisition-weighted stack of spirals (AWSOS) (11), 3D CONES (12), or Fermat-looped orthogonally-encoded trajectories (FLORET) (13). Concerning real spatial resolution and partial volume effects, these non-Cartesian techniques exhibit different behavior compared to conventional Cartesian sampling schemes. In addition, they are more susceptible to artifacts caused by B_0-inhomogeneity. On the other hand, non-Cartesian sampling schemes are well suited for the combination with iterative image reconstruction techniques. These issues will be discussed in the following five sections.

Real Spatial Resolution

The real spatial resolution of an imaging method refers to “the smallest resolvable distance between two different objects, or two different features of the same object (14)”. However, the definition of the “real spatial resolution” depends on the choice of the objects and also on the subjective perception of the observer. The spatial resolution in MRI is often defined via the highest k-space frequency (k_{max}) that is sampled. Under ideal conditions, this nominal spatial resolution (RES_{nom};

equation 1) equals the voxel size (Δx) of the Fourier transform. In an MRI experiment the real (or effective) spatial resolution is usually lower than the nominal spatial resolution.

$$RES_{nom} = 1/(2 \cdot k_{max}) \qquad \text{(equation 1)}$$

This loss in spatial resolution depends on the applied acquisition technique, the employed image reconstruction method, post-processing (e.g. filtering) and the transverse relaxation times of the investigated tissue. In addition, other factors such as patient motion can further reduce spatial resolution. An often employed measure that takes the aforementioned parameters (except for patient motion) into account is the point spread function (PSF). Generally speaking, the PSF describes the response of an imaging method to a point-like object. The image that is generated by an imaging method can be described as the convolution of the PSF with the real (i.e. ideal, perfect) image of the object. The full-width-at half-maximum (FWHM) of the PSF is often used as measure for the effective resolution.

In most non-Cartesian UTE techniques – that are used in X-nuclei MRI – a spherical k-space volume is sampled, whereas in Cartesian sampling schemes usually a cuboid is sampled. If other influences (e.g. T2* decay, filtering) are neglected, the PSF can be approximated by the Fourier transform of the sampled k-space volume. A spherical k-space volume yields a 1.3-fold broader FWHM of the PSF compared with a cubical k-space volume (1.59 pixels vs. 1.21 pixels). This signifies an increase in spatial resolution by a factor of approximately 1.3 when compared with spherical k-space sampling. Fast T2*-decay results in an additional increase of the FWHM. For conventional radial k-space trajectories with constant gradient strength the FWHM can be calculated analytically (15). For other trajectories such as DA-3D-PR and TPI the FWHM can be easily simulated (7). For common acquisition parameters and typical relaxation times, the FWHM is ≈ 1.6 – 2.0 pixels. If filters are applied, the FWHM is usually larger than 2.0 pixels. Considering the low spatial resolution of X-nuclei MRI (Δx ≈ 2 – 6 mm), this results in spillover effects that even influence the signal intensity of voxels that can be ≈ 1 cm away from a certain type of tissue. Thus, many applications require correction of these spillover effects.

Correction of Partial Volume Effects

Partial volume effects (PVC) can be caused by tissue fraction effects (i.e. one voxel contains two or more different types of tissue) or by signal spillover effects from neighboring voxels. This can result into a large bias in measured sodium concentrations. This is particularly pronounced for applications where large differences in signal intensities occur. For example in human brain, cerebrospinal fluid (CSF) exhibits a more than 3-fold higher sodium concentration than brain tissue. This might be also one of the causes for the large variations in measured tissue sodium concentrations that have been published. Literature values of sodium concentrations in brain white and gray matter range from 19-72 mmol/L and 30-62 mmol/L, respectively (16).

One method to reduce partial volume effects is the application of the geometric transfer matrix (GTM) method, which was developed for positron emission tomography (17). The GTM method requires accurate, high-resolution anatomical data and the knowledge of the PSF. Corrected signal intensities are derived for different regions of interest (e.g. white matter, gray matter, CS). It has been shown, that the GTM method can markedly improve the quantitative accuracy of sodium MRI of human brain (16).

Correction of B_0^-/ B_1^- Inhomogeneities

In non-Cartesian sampling schemes, off-resonances – that can be caused by B_0-inhomogeneities – lead to broadening of the PSF and to image blurring. For instance, a frequency segmented conjugate phase reconstruction can be used to reduce these artifacts (18). Correction requires a B_0-map that can be derived from images that have been acquired at two different echo times (either by ^{1}H or by ^{23}Na MRI).

The RF wavelength for most X-nuclei (except for ^{19}F) is more than a factor of two longer than the RF wavelength of ^{1}H. Thus, at currently used field strengths ($B_0 \leq 9.4$ T), whole-body X-nuclei MRI does not require parallel transmission capability. Nevertheless, B_1-inhomogeneities resulting from local X-nuclei RF coils can lead to a bias in quantitative X-nuclei MRI. There are several techniques available to correct this bias (18,19). If transceiver RF coils are employed, usually the principle of reciprocity is applied. It assumes that the transmit field (B_1^+) equals the receive field (B_1^-) (20). Common methods to map the B_1^+ field are the double-flip-angle method (DAM) (21), the phase-sensitive (PS) method (22) and the Bloch-Siegert shift (BSS). The DAM method calculates the effective flip angle from the ratio of two images that were acquired with different nominal flip angles of α_0 and $2\alpha_0$. The phase sensitive method encodes the magnitude of the B_1^+ field into the signal phase and requires a preparation pulse ($2\alpha_0$). The BSS method is also phase based and an off-resonant preparation pulse is used to encode the B_1^+ field (23). The phase-sensitive method has been shown to be more sensitive than the double-flip-angle (22) and the BSS method (19). However, the DAM method has the advantage that it requires only a conventional X-nuclei pulse sequence.

Reconstruction of Multi-Channel Data

Phased array receiver coils can be employed that make use of the increased sensitivity of small surface coil elements (3). The phased array technology is well established in ^{1}H MRI, however most X-nuclei MRI studies have been performed using single-channel RF coils. For ^{23}Na MRI of the human brain at 9.4 T, Shajan et al. presented a 27-channel ^{23}Na receive helmet, which is surrounded by a four-channel ^{23}Na transceiver array and a four-channel ^{1}H dipole array. At 7 Tesla, designs for a 15-channel (4) and a 30-channel array coil have been presented (5). Up-to-date overviews about high-performance RF coils for ^{23}Na MRI can be found in Wiggins et al. (6) and Bangerter et al. (7). However, for the combination of signal from different coil elements, the low SNR of X-nuclei MRI poses a different situation than in ^{1}H MRI. A simple sum-of-squares (SOS) reconstruction to combine the signals of each individual coil element can lead to severe noise amplification. Therefore, methods that exploit the knowledge of the sensitivity profiles of all coil elements in each voxel, such as the adaptive combination (ADC) algorithm (8,9) or the SENSEtivity (SENSE) encoding method (10), perform superior compared to a simple SOS reconstruction. In addition, the receive profile (B_1^-) of the array needs to be corrected for. For an array that is surrounded by a transceiver birdcage coil, the B_1^- field can be derived from the ratio of images that have been acquired by the array and the birdcage.

Iterative Image Reconstruction Techniques

Non-Cartesian k-space sampling schemes that are usually used in X-nuclei MRI, are well suited for the application of iterative reconstruction schemes. However, compressed sensing (CS) (17,18) and related iterative reconstruction algorithms are still rarely used in X-nuclei MRI (15,19,20), although they can markedly improve image quality. For example, 3D dictionary learning compressed sensing (3D-DLCS) enables precise reconstruction of undersampled ^{23}Na MRI data with markedly reduced

noise and artifact levels compared with conventional reconstruction (21). Also high-resolution anatomical information from ^{1}H MRI can be used to reduce partial volume effects and to increase the image quality in X-nuclei MRI. Even the most basic anatomical information – the shape of the object – can be incorporated into the iterative image reconstruction process to improve image quality (15). Information about tissue boundaries can further reduce image blurring and partial volume effects (16). However, in these iterative reconstruction techniques the PSF is locally dependent, which hampers the correction of remaining partial volume effects.

Summary

In X-nuclei MRI, the real spatial resolution is often lower than the nominal spatial resolution. In combination with large voxel sizes – that are required to achieve sufficient SNR – correction of partial volume effects are often required to reduce signal bias from neighboring tissues. In addition, correction of transmit and receive field inhomogeneities improve the quantitative accuracy. Iterative image reconstruction techniques can further improve image quality. To conclude, consideration of the discussed image reconstruction related issues can markedly improve the quantitative accuracy of X-nuclei MRI and has also the potential to reduce acquisition times.

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