Comparison of 3D Imaging Sequences for ^{23}Na MRI of In Vivo Kidney at 9.4 T

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INTRODUCTION:

Sodium (^{23}Na) MRI is a unique imaging modality giving essential information on cellular level, which may help to understand renal physiology, pathologies, as well as the pharmacological effect of drugs [1-4]. However, ^{23}Na MRI is challenging due to the low SNR, fast signal decay, and low gyromagnetic ratio, and it is therefore essential to make the right choice of the imaging sequence. The recently published ^{23}Na rodent or murine kidney studies have used Gradient Echo (GRE-3D) or Chemical Shift Imaging (CSI-3D) sequences, which however suffered from long TE or low SNR/time [1-3].

In this study we used an alternative sequence with anisotropic radial acquisition technique, Ultra-Short Time-to-Echo (3D-UTE), which allowed for TE down to 160 µs. These three sequences have been compared regarding effective spatial resolution in a resolution phantom, and regarding SNR/time in in vivo measurements. For a fair comparison, the parameters of the three imaging sequences were optimized to achieve an average scan time of about 11 min.

METHODS:

An in-house developed saddle-shaped **^{23}Na transceiver surface coil** was used in order to maximize the measured ^{23}Na signal intensity at 105.9 MHz in the kidney [5]. A custom-built linearly-polarized ^{1}H transceiver birdcage resonator was employed for ^{1}H reference images (Bruker BioSpin MRI GmbH). The effective spatial resolution variations due to partial volume effects were investigated using a **resolution phantom** with vials of different diameters in the range of 1 - 3 mm (see phantom layout in Fig. 1B - left). The resolution phantom was firstly filled with a 0.9% NaCl solution and secondly with a 0.9% NaCl gel with 3% agarose (Fig. 1A-B). The SNR/time in the renal tissue of a Wistar rat (300 g) using different imaging sequences was demonstrated by means of **SNR maps** (Fig. 1C), which were calculated as the signal intensity in each voxel subtracted by the mean noise and divided by the standard deviation of the noise. The MR images were carried out on a 9.4 T system (Biospec 94/20, Bruker BioSpin MRI GmbH, Ettlingen, Germany) with the following 3D pulse sequences: UTE-3D, CSI-3D and GRE-3D.

The **UTE-3D** sequence, which is an FID readout technique [6], was used with an anisotropic sampling strategy. The reduced sampling points spacing in z-direction (Δkz) enabled increased SNR/time in axial sections. UTE-3D allowed for very short Echo Time (TE) down to 160 µs, which was strongly dependent on the available transmit power for the resonator system, i.e. the duration of the transmit pulse, since TE was defined as the time period from the middle of the non-selective RF excitation pulse to the beginning of the acquisition window. The **CSI-3D** pulse sequence basing on the pixel-by-pixel FID readout used a weighted acquisition technique which averaged the number of scans at certain phase encoding steps via Hanning function. The nominal spatial resolution (FOV/nr. of phase enc. steps) was defined as the 64% width of the point spread function. Therefore, the hanning-weighted CSI-3D sequence needed more phase encoding steps to obtain the same spatial resolution as the standard CSI-3D sequence. In order to make UTE-3D and CSI-3D scan times comparable, the number of scans in CSI (40000) should be matched to the number of acquired data points (nr. of half-projection · nr. of phase enc. steps=12753·32=408096) in UTE. Thus, spectroscopic CSI-3D data was averaged over 10 FID data points. Furthermore, the axial FOV of CSI-3D was adapted from 64^2 to 45^2 mm^2, to reduce the handicap of time consuming pixel-by-pixel acquisition in CSI-3D – a fair limitation to acquire images with comparable noise contribution. The **standard GRE-3D** sequence used in this study was basing on a Cartesian sampling strategy and on the asymmetric echo formation (10%) to fasten the acquisition. Further parameters like BW, resolution, transmit pulse, FOV and matrix size were the same as in the UTE-3D sequence (for imaging parameters please see Table 1). The ^{1}H MR images with high T2-contrast were recorded using a RAREst sequence with TE / TR = 3 ms / 30 s, RARE factor of 36, FOV of 64^2. Twenty axial slices were acquired with the slice thickness of 1 mm and an in plane resolution of (0.5 x 0.5) mm^2 in a scan time of 60 sec. The ^{1}H and ^{23}Na MR images shown in Fig. 1 are zero-filled by factor 2.

RESULTS: The comparison of three imaging sequences in resolution phantom (Fig. 1A) and in in vivo kidney measurements (Fig. 1B) showed that anisotropic 3D-UTE allowed for the highest SNR/time in the renal tissue (25.3±0.02), and the 3D-UTE sequence outperformed the GRE-3D and CSI-3D sequences by factor >2. However, UTE-3D suffered from effective spatial resolution loss (pixel deviation by factor 1.5) compared to CSI-3D.

DISCUSSION AND CONCLUSION: The GRE-3D images are dominated by the T2* signal losses (comp. Fig. 1A and B) caused by long TE > 2 ms. High (1x1) mm² in-plane resolution could be achieved in recent in vivo rodent kidney studies [1,2,4] through anisotropic sampling schemes that were also implemented for UTE-3D in this work. The anisotropic sampling in radial acquisition could allow for SNR/time as high as the isotropic density-weighted radial acquisition technique [7]. The non-Cartesian radial UTE-3D sequence proved to be less prone to the movement and to the T2* signal losses. This fast acquisition could also benefit further X-nuclei MRI with short T2 decaying times (e.g. 39K, 35Cl). Much shorter TEs are possible by using e.g. ZTE imaging sequence [8]. To achieve higher SNR/time with combination of raw data undersampling, alternative gradient encoding possibilities can be employed, e.g. spiral or twisted projection techniques [9]. However, the spatial resolution loss of non-Cartesian sequences still remains, whereas the pixel deviation of UTE-3D was in the range of 1.5 to 1.7 pixels and in good agreement with the theoretical investigations published before [6,7]. The CSI-3D sequence is more convenient for measurements including multiple averages, which however allowed accurate resolutions without pixel deviation. Long measurement times (Tacq ≈ 50min) are required when images without T1 weighting should be acquired, e.g. with TR>150ms for the quantification of sodium concentration. CSI-3D was experienced to be less affected by the motion due to the Hanning function weighting. CSI-3D sequence can be further optimized by adaptation of phase encoding gradients in order to shorten encoding gradient duration before acquisition, especially relevant for data sampling in the k-space center. The spatially-encoded spectroscopic information gained by the CSI-3D sequence remains as a major advantage compared to other sequences.

REFERENCES: [1] Maril et al., Kidney Int. 69:765-768 (2006); [2] Maril et al., Kidney Int. 65:927-935 (2004); [3] Neuberger et al., MRM 58(5):1067-71 (2007); [4] Atthe et al., Am J Physiol Renal Physiol 297 (2009); [5] Kalayciyan et al., ISMRM, #337, 2012; [6] Rahmer et al., MRM 55:1075-1082 (2006); [7] Nagel *et al.*, MRM 62, 1565-1573 (2009); [8] Weiger et al., Enc MR (2012); [9] Boada et al., MRM 38, 1022-1028 (1997).

Acknowledgment: This work was financially supported by Friedrich Naumann Foundation. Special thanks to the Institute for Embedded Systems, HS Mannheim.

Sequence	UTE-3D	GRE-3D	CSI-3D
TE/acq. delay	165 µs	3 ms	acq.del.=0.49 ms
TR	50 ms	20 ms	20 ms
Tacq	11 min	11 min	13min
Matrix	64^3	64^3	45x45x32
BW	5 kHz		
nominal resol.	(1x1x4) mm^3		
Excit. pulse	3ms block pulse / 16.7 dB		
SNR	25.3±0.02	12.4±0.04	11.5±0.03

Tab. 1 – Overview of the imaging sequence parameters and the SNR in the renal tissue.

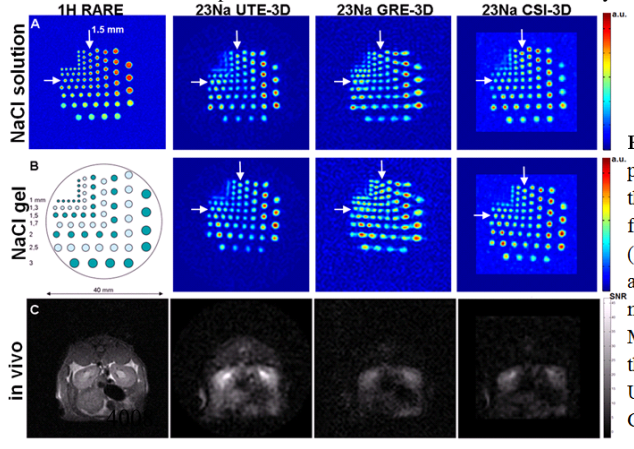

Fig. 1 – Comparison of partial volume effect in the resolution phantom filled with NaCl solution (A) and NaCl gel with 3% agarose (B), and SNR-maps of in vivo kidney MR measurements for three imaging sequences: UTE-3D, GRE-3D, and CSI-3D.