BILATERAL KIDNEY ^{23}NA-MRI: QUANTIFICATION OF TISSUE SODIUM CONCENTRATION BY USING A TWO-ELEMENT PHASED ARRAY SYSTEM WITH HOMOGENEOUS B1-FIELD EXCITATION AND ULTRA-SHORT TE

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Introduction:

Tissue Sodium Concentration (TSC) has been suggested as a measure for renal function in *in vivo* models of acute tubular necrosis and pharmacologically-triggered diuresis [1, 2, 3]. Besides homogeneous transmit-B_1-field and excellent receiver sensitivity, TSC quantification requires ultra-short echo times and long TR. Nevertheless, earlier *in vivo* studies measured TSC from image data which was acquired with relatively long TE and short TR [1, 3]. Despite relaxation time corrections these were applied as a temporally-independent constant, although T_1 and T_2* relaxation times may change during cortical ischemia and diuresis – an effect which can result in large variations of the estimated TSC.

The aim of this work was to develop an RF resonator system and to adapt an Ultra-Short Time-to-Echo (UTE) sequence for the quantification of renal TSC in renal *in vivo* models of common human diseases. In order to maximize the ^{23}Na receiver sensitivity a 2-element phased-array was combined with a commercial double-tuned (^{23}Na/^{1}H) birdcage volume resonator forming a dual RF resonator system. The quantification method was then applied to measure TSC changes in rat kidney before and after diuretic agent injection.

Methods:

Two ellipsoidal surface coils (33mm x 23mm) were designed to cover one of the two kidneys, respectively. The developed ^{23}Na receive-only surface coil was noise-matched via low noise preamplifiers (LNPs) introduced by Roemer *et al.*, which is a commonly used concept for decoupling multiple receiver elements in phased array coils [4]. The LNPs (MwT, USA, 2.2 Ω input impedance, 28.8 dB gain, 0.45 noise figure) were noise-matched to the detector element via a λ/4 transformation network [5], which was modified to minimize the number of electronic circuit components. During the design the transformation path of the network between coil terminal and preamplifier was adjusted to the resonance frequency of 105 MHz. The efficiency of this trap circuit was measured via a S_{21} measurement with decoupled probe pair (0.5 cm Ø) brought to near proximity of the receive-only coil: The peak split was measured to be 20 MHz and -35 dB preamplifier decoupling was achieved. The decoupling between transmit and receive elements was achieved in a first step by arrangement of the birdcage resonator's B_1-field orthogonal to the surface coil's B_1-field. In a second step, an active decoupling unit was integrated using a PIN-diode, which was controlled in active low mode by DC power supply through RF interface (U_{TRA} =+5V during transmission, and U_{REC} = -34V during acquisition). The LNP was supplied by a 10 V DC voltage through a stabilizer. The double-tuned (^{1}H/^{23}Na) linearly-polarized birdcage coil (Bruker Biospin MRI GmbH, Ettlingen, Germany) and the 2-channel noise-matched phased array system are presented in Figure 1a and 1b, respectively.

In order to quantify the TSC, firstly surface coil's receive sensitivity was compensated for through a reference scan method in a homogeneous reference phantom [6]. Secondly, sample and reference scans were co-registered using the position of two fiducial vials (155 mMol NaCl) permanently fixed on top of the receiver coil (see Fig. 1b). Lastly, the sensitivity corrected ^{23}Na image was computed as the quotient image from the sample and the reference scans, multiplied by a correction factor for different coil loading. The quality of TSC quantification of the newly-developed resonator system was proven on a diuretics-induced kidney model. Renal ^{23}Na signal changes in a healthy female rat (~280 g) were observed 30 min prior to the furosemide injection (10 ml/kg body weight), and up to an hour after bolus injection. 3D-UTE pulse sequence allowed for short TE_{UTE} = 185 µs, and long TR = 150 ms for nominal voxel size of 1x1x4mm^3 in T_{acq} = 10 min. Further parameters were: 4206 projections, BW = 5 kHz, 0.35 ms block pulse length, and FOV = (6.4x6.4x6.4) cm^3. ^{1}H images co-registered with ^{23}Na images before and after furosemide injection can be seen in Fig. 2 d, e. The acquired sodium data from each channel was reconstructed with sum of squares method to compose the final ^{23}Na-MR images. ^{1}H-reference scan was recorded with ^{23}Na/^{1}H transceiver volume resonator by means of a 3D-RARE sequence using TE/TR = 3 / 10 ms, and a resolution of (0.5 x 0.5 x 2) mm^3 in 10 sec (respiratory triggering time excluded) on a 94/30 Bruker Biospec system (Bruker BioSpin GmbH, Ettlingen, Germany) under permission and approval of the local animal welfare committee.

Results and Discussion:

Different renal subregions such as the medulla and cortex are clearly delineated by high T_2-contrast achieved in transaxial ^{1}H image (Fig. 2a). The single coil ^{23}Na images are presented in Fig. 2b and 2c. The sum-of-square image in Fig. 2d demonstrates the excellent spatial resolution and high sensitivity of the developed bilateral kidney imaging system. A dual resonator system allows the design of two receiver surface coils, so that each one can be localized directly above each kidney. This proved to be an advantageous concept for bilateral renal imaging compared to transceiver surface coil, which requires larger surface coil in order to observe both kidneys. A clear decrease in density-weighted ^{23}Na signal was measured after furosemid injection (Fig. 2e). Furthermore, since extremely homogeneous B_1-field was generated by the large volume resonator (<5% B_1-field variations across the abdominal range), no B_1-field corrections were necessary for TSC quantification. The TSC maps for each kidney before and after furosemid injection is shown in Fig. 3. The TSC decreased from 180±10mmol/l to 100±10mmol/l in the inner medulla and slight increase of the TSC from 60±10mmol/l to 90±10mmol/l was measured in cortex. Others determined max TSC to be 300mM [2], which may be higher due to differences in the investigated kidney physiology, e.g. different initial water conditions in the sample. Yet, the quantification accuracy in [2] remains to be questioned since long TE and short TR, but static relaxation time correction was applied. On the other hand, the quantification accuracy of herein applied study has been previously demonstrated to be better than 10mmol/l [7, 8]. In conclusion, the newly-developed resonator allowed observing changes in TSC values with high sensitivity. Improved hardware and optimized 3D-UTE pulse sequence allowed for precise TSC quantification, which will be applied to study TSC in various renal pathologies in the future.

References: [1] Maril et al., Kidney Int. 69:765-768 (2006); [2] Maril et al., Kidney Int. 65:927-935 (2004); [3] Atthe et al., Am J Physiol Renal Physiol 297 (2009); [4] Roemer et al., MRM, 16(2):192-225 (1990); [5] Reykowski et al.,MRM, 33(6):848-852 (1995); [6] Ouwerkerk et al., Breast Cancer Res Treat. 106 (2007); [7] Kalayciyan et al., Proc. ISMRM 1891 (2010); [8] Wetterling et al., Phys. Med. Biol. **55** (2010) 7681–7695.

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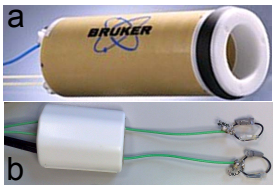

Fig. 1 – Dual RF resonator system composed of (a) ^{1}H birdcage volume resonator, and (b) 2-element receiver.

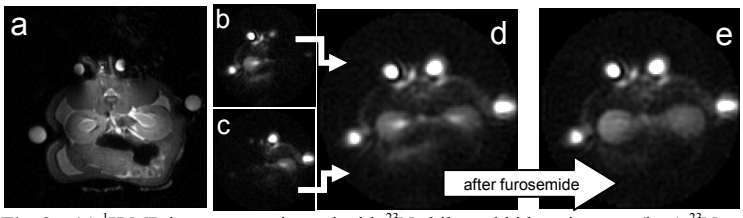

Fig. 2 – (a) ^{1}H MR image co-registered with ^{23}Na bilateral kidney images. (b, c) ^{23}Na single-coil images acquired by phased array coil are depicted as (d) sum-of-squares image. (d ,e) ^{23}Na images before and after furosemide injection are shown.

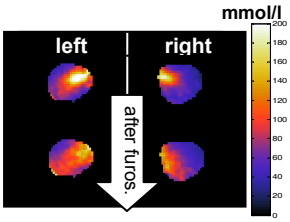

Fig. 3 – TSC maps of the left and right kidney before (top row) and after (bottom row) furosemide injection.