A Double-Tuned ^{1}H/^{23}Na Two Element Phased Array System for In Vivo ^{23}Na Magnetic Resonance Microscopy at 7 T

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INTRODUCTION: There is growing interest in and evidence for the use of local Tissue Sodium Concentration (TSC) levels as biomarkers for tissue viability and integrity in stroke [1] and tumor [2]. However, the SNR in ^{23}Na Magnetic Resonance Microscopy (^{23}Na-MRM) images is low due to the ^{23}Na-nuclei's low *in vivo* concentration, low gyromagnetic ratio, fast signal decay, and the required small voxel sizes (< 4 μl) in rodent brain imaging. The aim of this study was to develop a two-element phased array system in order to maximize the SNR in the ^{23}Na channel while also allowing for the acquisition of high-resolution anatomical ^{1}H MRM images without the need to change the coil system during the experiment.

METHODS: Preamplifier matching is a well known method to suppress cross-talk in multi-element receiver coils [3]. In order to achieve optimal signal transmission from the RF coil to the MRI system's preamplifier, the RF detector element impedance must be precisely matched to the preamplifier's input impedance (typically 50 Ω), which requires a λ/4 transformation circuit. At 79.4 MHz (for ^{23}Na at 7 T), the use of a relatively long λ/4-cable is cumbersome (λ/4 ~ 64 cm at 79.4 MHz) and thus the transformation circuit must be replicated by a pi-network as described elsewhere [4]. However, for small detector sizes (< 20mm inner diameter) and using a commercially available low input impedance preamplifier (MwT, USA, 2.5 Ω input impedance, 26 dB gain, 0.5 noise figure), the coil decoupling achieved with such a pi-network was suboptimal as measured by the low peak split (< 12 MHz) and the low s_{21}-transmit current suppression (-10 dB). The low inductive and thus low capacitive detector element impedances may have reduced the trap circuit efficiency of the preamplifier path. In order to improve the coil decoupling, the coil element impedances were optimised by double-winding the detector element in order to increase the inductive and capacitive detector element impedances. By employing two windings in series (1 square loop with 26 mm side length and 1 circular loop with 20 mm diameter), the coil capacitance was decreased from 180 pF (11.2 Ω) to 85 pF (23.7 Ω). The peak split, measured to be 24 MHz, was doubled by a factor of two for this newly-developed coil shown in Figure 1. The coil elements were anatomically shaped by bending them on a 42 mm diameter cylindrical former. The preamplifier decoupling was improved to < - 16 dB. Furthermore, a ^{23}Na/^{1}H birdcage resonator with 72 mm inner diameter (i.d.) was developed to maximise the transmit B_1-field homogeneity and hence the detected SNR, to allow for accurate quantitative ^{23}Na-MRM together with the acquisition of high resolution anatomical ^{1}H images. The ^{23}Na birdcage channel could be actively detuned from the ^{23}Na phased array coil during the receive mode. In addition, passive decoupling circuits were inbuilt on the phased array elements to decouple them from the birdcage resonator during the transmit mode. The ^{23}Na-imaging capability of the two-element phased array coil was compared to a commercially available double-tuned ^{23}Na/^{1}H planar transceiver surface coil (Bruker BioSpin, Ettlingen, Germany), by imaging a 30 mm tube filled with non-physiological 1M NaCl solution using a 2D-FLASH sequence on a Bruker BioSpec 7 T system with TE/TR = 5/250 ms, voxel size = 1.25 x 1.25 x 2 mm^3, and an acquisition time of 10 min, as presented in Figure 2. To provide a fair comparison of each coils' performance at depths typical of a rat's brain, the 90° flip angle for the transceiver coil was set at a depth of 12 mm. As a consequence, saturation bands corresponding to flip angles > 90° are evident at depths < 12mm for the transceiver coil, with lower flip angles occurring for depths > 12 mm. The SNRs as a function of depth are compared in Figure 3 using the average of five vertical profiles taken across the respective SNR parameter maps. To demonstrate the applicability of the developed phased array resonator system, *in vivo* ^{1}H and ^{23}Na MRM images of a healthy adult rat (~ 400 g) were acquired as follows: ^{23}Na - 3D-FLASH sequence with TE/TR = 2.5/23 ms, voxel resolution = 1 x 1 x 4 mm^3, and an acquisition time of 40 min; ^{1}H - 2D RARE sequence with RARE factor 8, 234 x 234 μm^2 in-plane resolution and 2 mm slice thickness, TE/TR = 33/2500 ms, and an acquisition time of 5 min. .

RESULTS and DISCUSSION: An SNR improvement of 25 % in favour of the phased array coil was measured at a depth of 12 mm, which most likely derived from its optimized and more anatomically-shaped detector elements. The *in vivo* ^{1}H and ^{23}Na MRM images for the two-element phased array coil are shown in Figure 4; SNRs measured in the ^{23}Na images were 10 in the brain and 30 in the ventricular cerebrospinal fluid. To compare these ventricular SNR measurements with comparable measurements in the literature, an 'equivalent SNR' was calculated, correcting for field strength, acquisition time and voxel resolution differences used in the reported studies (i.e. commensurate with all images being acquired at 7 T in 40 minutes with 4 μl voxel resolution). An SNR improvement of 42 % was measured for the two-element phased array in the current study compared to the next-best coil, a double-tuned ^{23}Na/^{1}H transceiver surface coil developed by Alecci *et al*. [5]. , with a six-fold improvement compared to a more recently published ^{23}Na MRM study using a novel triple resonant ^{1}H/^{23}Na/^{39}K transceiver coil system [6]. The high SNR measured for the developed coil system coil reflects the improved coil design. Using this newly-developed coil, *in vivo* rat brain ^{23}Na-MRM images with high spatial resolution can be acquired in a reasonable scanning time. In conclusion, the current study provides improved spatio-temporal resolution coupled with the ability to quantify measures of Tissue Sodium Concentration, allowing for the assessment of tissue viability in acute stroke tissue.
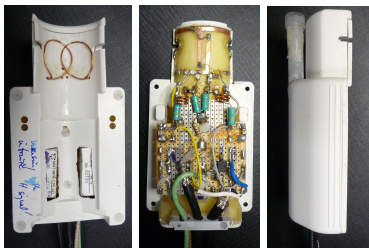

Figure 1: Newly-Developed ^{23}Na receive-only two-element phased array coil in (left) bottom, (middle) top), and (right) side view
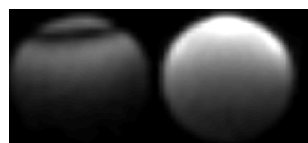
Figure 2: NSR maps as computed from the ^{23}Na MRM images of a homogeneous phantom acquired with (left) transceiver surface coil and (right) two-element phased array. The SNR grey scale levels are the same in each image.
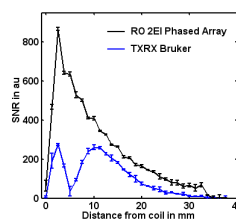
Figure 3: SNR as a function of depth for the two-element phased array and the commercial transceiver surface coil.
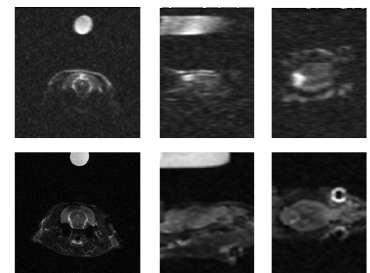
Figure 4: axial, coronal, and sagittal ^{23}Na (upper row) and ^{1}H (lower row) MRM images.

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