Investigation of Flexible Transmit/Receive Coil Concepts on B1+ Performance at 3T

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Introduction

A significant reduction in RF power, a decrease in SAR, an improved compatibility with implants and a more distinct control of the B_1^+ field distribution are attractive motivators for local transmit and receive (TX/RX) arrays when compared with large volume body coils. Local transmit coil arrays have shown to offer benefits and improvements for cardiac and body applications with reasonable B_1^+ homogeneity and acceptable SAR at 7 T [1,2]. These coil concepts also have been migrated to 3 T [3,4] in order to provide a further reduction in SAR, in particular when combined with SAR-exact B_1^+ maps [5]. In contrast, RF transmission using large, system-integrated whole body coils can result in insufficient B_1^+ homogeneity or increased scan time. For whole body applications, larger arrays with a higher number of individual elements are desired, which cover the total required FOV and better serve clinical needs and patient comfort. Transmit arrays and their applications must be robust in positioning, transmit homogeneity, coil matching, and required RF power. In this study, we investigate the influence of coil bending of flexible transmit arrays on homogeneity, required RF power and SAR. We present B_1^+-shimming using flexible 8-channel local TX/RX arrays for the human torso at 3T [6,7] using fast B_1^+ mapping by means of DREAM [8]. B_1^+ mapping is carried out for two different coil array designs. Both arrays do have the same overall geometrical dimensions and cover the same FOV, but differ in individual coil size, distance and decoupling. Using the two arrays, different concepts of flexible TX/RX coils for functional and anatomical whole body imaging are investigated.

Methods

Two different coil arrays were built up, both with a length of 180 mm in z- direction, using a strip conductor width of 7mm. The first coil design ("gapped") has individual coil elements (w=105mm), separated by a gap of 18 mm, (between conductor centers). Mutual decoupling is performed using local inductive transformers. The second coil ("overlapped") has larger coil elements (w=135mm), and decoupling of adjacent elements is achieved by overlap. Each element of both coils is split six times and resonates in connection with ATC 100C and Voltronics tuning capacitors (NMAJ20-4). The coils are matched to a phantom load using serial inductors. Matching was performed for 30 mm distance between phantom and coil. Overlap and components such as decoupling transformers and capacitors were determined using the EM simulation tool HFSS [9]. While the elements in the center are not influenced much, return loss of the outer coils is influenced by different loading conditions. The measured RF power is corrected by the transmission losses of the RF chain and the individual return loss of the coil elements. Common mode currents are suppressed using cable traps directly downstream of the matching circuit, providing stable tuning and decoupling of the individual elements.

Experiments were performed on a modified 3T Philips Achieva MRI scanner with 8 independent TX channels [10,11]. In a first set-up, the coil arrays are placed in a planar shape. 2D B_1^+ maps of the individual array elements are acquired in a dynamic loop using DREAM (transverse orientation, field-of-view (FOV) = 450×270 mm^2, scan matrix= 128×40, phase-encoding direction: AP, slice thickness = 10 mm, STEAM flip angle α= 60°, imaging flip angle β= 10°, TR= 4 ms, TE = 1.2 / 2.3 ms, shot duration = 170 ms, delay between successive shots = 1 s, total scan duration for 8 TX channels = 9 s. The magnitude B_1^+ maps are used for B_1^+ shimming incorporating the phases of the underlying source images to achieve a homogeneous excitation in the intended imaging slice. Finally, the shimming experiment was repeated with both coils closely wrapped around the phantom maintaining the minimum distance of 3cm.

Results

Decoupling of the unloaded coil elements is -25 dB and -15 dB for the loaded case (torso shaped water phantom (350*350*190mm), filled with 9,6l H2O, 14,4kg $C_6H_{12}O_6$, 0,96kg NaCl. Fixed RF power matching of the individual elements results in a worst case matching of -8dB for the outer coil elements in the planar configuration, compared with better -15 dB for the center coil elements. Slightly better homogeneity for both coil designs is achieved, when the coils are bent around the phantom. Total required RF power for 10uT was measured to be < 2kW for both coil designs. EM simulations and measurements show better B_1^+ homogeneity of gapped coil arrays compared with the overlapped coil array (Fig.1). In vivo Image of a transversal and coronal slice through the abdomen (FOV = 450x270 mm², pixel size = 1.4x1.4mm², slice thickness = 10mm, TR/TE = 7/2.3 ms, flip angle = 20 deg, peak B_1 = 5 uT, scan time = 1.3 s) is shown in Fig.2

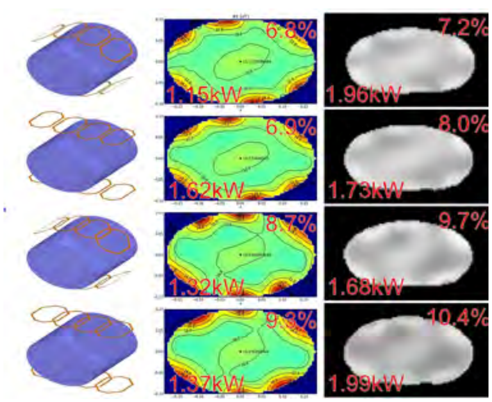

Fig. 1: Geometric coil models (left), EM simulated (center), and experimentally measured (right) at 3 cm distance from phantom. B_1^+ maps of an elliptical water phantom acquired using DREAM at B_1^+=10µT. Different shim settings are applied for each coil. Gapped coil arrays provide better homogeneity (cv%) compared with overlapped coil array.

Fig. 2: In vivo abdominal images (right) of transversal, coronal and sagittal slices through the abdomen. For quadrature excitation, TX signal voids occur (left, labeled by arrows), which are removed for an RF shimmed excitation (right). For signal reception, a sum-of-magnitude based coil superposition was employed (FFE, multi-2D, FOV = 450x270 mm², pixel size = 1.4x1.4mm², slice thickness = 10mm, TR/TE = 7/2.3 ms, flip angle = 20 deg, peak B_1 = 5 uT, scan time = 1.3 s per slice). For quadrature excitation, two pencil beam shaped TX signal voids are observed in feet-head direction, which are removed by an RF shimmed excitation.

Discussion

The presented approach using flexible transmit coil arrays shows the robustness of the proposed concept regarding homogeneity and RF power matching with respect to coil positioning and coil bending, which is an important clinical criterion. The coil design can significantly lower the RF power consumption and delivers better homogeneity results compared with a large system body coil at 3 T. Flexible TX/RX coil arrays may impact the RF frontend regarding improved efficiency, SAR constraints, number of TX-channels, and shortening of RF pulses, TE and TR compared with conventional system-integrated body coils for efficient inner volume MRI scans.

References

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