$B_1^{(+)}$ steering by an adaptive 4-channel transmit/receive coil array

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Introduction

Due to the wave behavior of the B1 field inside a body-sized object at B0 fields ≥ 3T, the clockwise rotating transversal B1 field component $B_1^{(+)}$ may be actively controlled by proper adjustments of the transmitting amplitudes and phases applied to a multi-channel transmit coil array. In this way adaptive coil control schemes [1] like B1-shimming, "switched" B1-gradients, transmit SENSE and RF-controlled volume selective preparation of magnetization may be achieved. We demonstrate the feasibility of this $B_1^{(+)}$ steering scheme in phantoms and in vivo by measurements using a 4-channel current sheet antenna (CSA) transmit/receive array together with a 3T MR scanner equipped with 4 fully independent transmit channels.

Methods

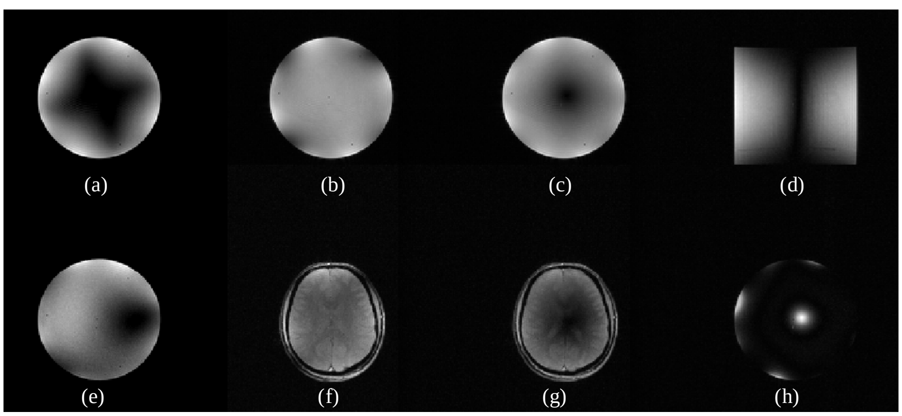
All measurements were performed on a 3T MR scanner (MEDSPEC30/100, Bruker Biospin MRI) equipped with four 4KW transmit channels capable of generating 4 independent RF pulses with individual sequence timing, pulse shape, frequency, phase and amplitude. Actively driven PIN diode T/R-switches and 4 high power circulators ensure a well defined termination of each coil element both during transmission and receiving. The coil array, consisting of 4 CSA's, was loaded either by a cylindrical phantom (i.d. 20 cm, $\varepsilon = 76$, $\sigma = 0.33$ S/m) or by the head of a healthy volunteer. All images were taken by a conventional low flip angle gradient echo sequence with amplitudes and phases of all RF pulses set separately. Images were reconstructed by the sum-of-squares-method and hence the detection sensitivity $|B_1^{(-)}|$ is rather homogenous over the field of view. Thus, low flip angle images reflect roughly the $B_1^{(+)}$ distribution. To demonstrate a volume selective preparation of magnetization a 1ms rectangular pulse preceding the imaging sequence was used to obtain a RF-controlled saturation pattern. Numerical calculations utilizing the XFDTD software package (REMCOM Inc.) were performed to predict $B_1^{(+)}$ distributions and the required amplitude and phase values. The numerical model, consisting of the CSA array and the phantom as load, was implemented on a 0.5 cm grid (49x61x49) with 8 perfectly matched layers. For simulation one coil of the array was excited by a CW RF current, applied at the matched feeding port, to infer the steady state field values. Because of symmetry the RF-field distributions of the remaining 3 coils were generated by 90°, 180° and 270° rotations of the calculated B1 field values.

Results

In Fig.1a-h the steering capability of the adaptive 4-channel transmit/receive coil array is illustrated for several RF driving conditions. Fig.1a reflects the "wrong rotating case" (0°, 270°, 180°, 90°) whereas Fig.1b shows the desired homogenous excitation pattern (0°, 90°, 180°, 270°). The "donut" distribution (Fig.1c) was obtained by driving the 4 coils with phases 0°, 180°, 0°, 180° respectively. Fig.1d is the corresponding coronal view of Fig.1c. Measured (Fig.1a-c) and simulated (Fig.2a-c) $B_1^{(+)}$distributions agree well. Fig.1e reflects the measured $B_1^{(+)}$ distribution when exciting a single coil at the top position clearly illustrating the chirality of the NMR active RF magnetic field component. Fig.1f,g illustrate in vivo images of the head obtained with homogenous and "donut" excitations which are very similar to corresponding phantom images. Fig.1h demonstrates a volume selective preparation of magnetization using a RF pulse with "donut" distribution for saturation and a homogenous pattern for the imaging RF pulse. The artifacts at the boundary of the phantom can be avoided by using an adiabatic saturation pulse.

Conclusion

Using a 4-channel CSA transmit/receive coil array we demonstrated the feasibility of $B_1^{(+)}$ steering using 4 fully independent transmit channels of our 3T MR scanner. In this way adaptive coil control schemes like B1-shimming, "switched" B1-gradients, transmit SENSE and RF-controlled volume selective preparation of magnetization may be utilized in high and ultra-high field MRI. Accurate FDTD simulations are necessary for determining the driving parameters (RF amplitudes and phases) for the desired B1-distribution.



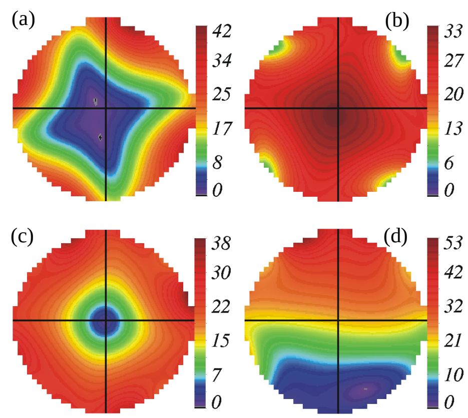


Fig.1: (a-c, e-h) Measured central axial images (cylindrical phantom, human head) for different excitation patterns, see text; (d) coronal view of (c)

Fig.2: $B_1^{(+)}$ distributions in μT (1kW total driving power) calculated by FDTD for the "wrong rotating case" (a), homogenous (b) and "donut" excitation (c). A nearly linear vertical B1-gradient (d) is obtained by using phases of 0°, 82°, 229° and 70° and amplitudes of 167W, 421W, 285W and 127W.

References:

1. F. Seifert et al., Proc. Intl. Soc. Mag. Reson. Med 10 (2002) 162