Active digital decoupling for multi-channel transmit MRI Systems

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Introduction: Methods like RF shimming [1] were proposed to compensate for wave propagation effects occurring in patients at field strength of 3T and above. Also, spatially selective excitation [2] in combination with Transmit Sense [3] can be applied to balance signal variations. The implementation of these methods requires several independent RF transmit coil elements around the imaging volume. The undesired effect of mutual coupling of RF coils can be compensated with passive [4] or active [5] methods. This work describes another active approach to control the currents in the transmit coils in order to obtain independent transmit sensitivities. The mutual coupling between the channels is compensated by a proper adjustment of the amplitude and phase of the input signals in the digital domain. In a first step, the coupling parameters of the transmit coils are determined via a set of preparation measurements. The results of these measurements are processed to calculate the required compensation parameters to correct the input signals for a desired target distribution of the currents in the transmit elements. The proposed method offers precise knowledge of the currents in the elements, which is relevant for the safety of the patient (SAR) and the device. Furthermore, it avoids possible errors in the determined sensitivities arising from large flip angle effects or the said wave propagation effects.

Theory: An array of RF coil elements can be described by its port impedance matrix $\underline{Z}_{PP}$ and its port equation $\underline{V}_P = \underline{Z}_{PP}\underline{I}_P$ with voltages $\underline{V}_P$ and currents $\underline{I}_P$ at the ports. The relationship between the voltages at the input ports and the currents in the transmit coil conductors can be described by another impedance matrix $\underline{Z}_{PT}$

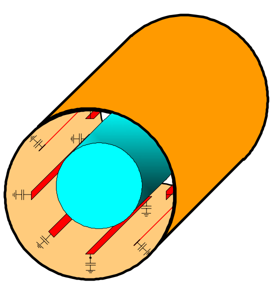
$$\underline{V}_P = \underline{Z}_{PT}\underline{I}_T \text{ (Equ.1)}$$

where the voltages $\underline{V}_P$ are produced by independent transmit signal sources at the ports, and the currents $\underline{I}_T$ in the transmit coils produce a target magnetic field $\underline{B}_1$ for spin excitation. The matrices are influenced by the load of the transmit coils. The inverse of the impedance matrix $\underline{Z}_{PT}$ can be determined by exciting the input ports by a set of linearly independent input signals $\underline{V}_P$ and measuring the currents $\underline{I}_T$ for each element of the set. Once the $\underline{Z}_{PT}$ matrix is known, the required input signals $\underline{V}_P = \underline{V}_{required}$ can be calculated for a target current distribution $\underline{I}_T=\underline{I}_{target}$.

Methods: The proposed method was verified by simulation and measurements using a whole body 3T MRI system extended to eight parallel RF transmit channels (Achieva,Philips Medical Systems, The Netherlands) [6]. An eight-channel TEM resonator head coil (Fig.1) was modelled using the Method of Moments software package FEKO [6]. Each TEM element consists of a 28x1cm copper strip at a distance of 1cm above a cylindrical RF screen (∅=36cm). A cylindrical phantom (∅=20cm, σ=0.5S/m, ε_r=80) was used to model an appropriate load. Tuning and matching were performed to obtain the homogenous mode of the TEM resonator with the current distribution $I=I_0\cos((n-1)2\pi/N)$ at the Larmor frequency f_L. The matrix $\underline{Z}_{PT}$ was determined using FEKO and (Equ.1). The required voltages $\underline{V}_{req}$ were calculated for an exemplary current distribution $\underline{I}_T=\underline{I}_{target}$=(1,0,…, 0) at f_L. The effect of load movement was tested by applying the formerly derived $\underline{V}_{req}$ for a 2cm radial shift of the phantom. Subsequently, the principle was demonstrated using a cylindrical eight-channel TEM resonator (Fig.2) in an MRI experiment. The tuning and matching and the dimensions of the resonator and the load are comparable to those of the simulation. In the first imaging experiment, only one of the channels was excited. In a second step, the measured coupling matrix $\underline{Z}_{PT}$ of the eight-channel system was used to compensate for mutual coupling. The resulting images (FFE, 6.2ms TR= 12.4ms, α=5°) of an MRI phantom (mineral oil, to demonstrate the method without wave propagation effects) were compared to evaluate the decoupling method.

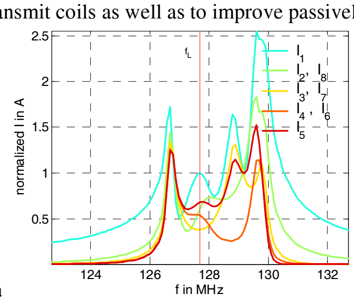
Results / Discussion: The results of the simulation are shown in Fig.3. Fig.3a shows the current distribution of the coupled TEM resonator with the independent element currents driven at element one. The corrected input signals $\underline{V}_{req}$ derived from the $\underline{Z}_{PT}$ matrix are applied to obtain the currents in Fig.3b. Although the $\underline{Z}_{PT}$ was determined at f_L, the target current distribution $\underline{I}_{target}$ is verified within a sufficient bandwidth for MRI. Within this bandwidth variation of the currents (Fig.3c) in the elements due to the 2cm shift of the phantom is negligible. The results of the MRI experiment are shown in Fig.4. Driving only one element (Fig.4a) results in the expected homogenous excitation of the phantom indicating strong coupling between the elements. Fig.4b demonstrates the result of the compensation method. The current in only one of the eight elements results in the required gradient shaped excitation profile.

Conclusion: The proposed method was successfully demonstrated in simulation and experiment indicating the potential of the method. The principle may be used to decouple arbitrarily shaped or flexible transmit coils as well as to improve passively decoupled multi-element coils.

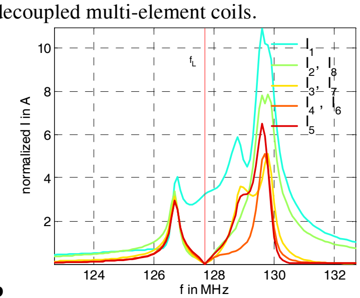


1 Sketch of the head coil model comprising the RF screen, the TEM elements and the phantom.

2 Photo of the eight-element coil loaded with the oil phantom.

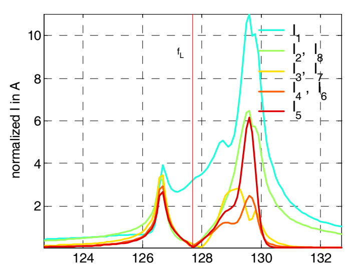


3a



3b

3 Simulated distribution of the normalized currents in the head coil elements **a**) uncompensated **b**) compensated by using active decoupling to obtain the target current distribution $\underline{I}_T$ = (1, 0,…, 0) and **c**) after shifting the load by 2cm in radial direction.



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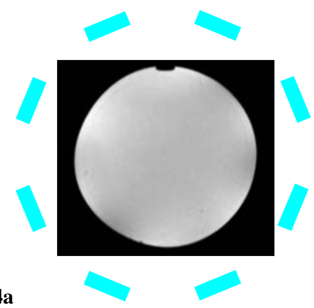
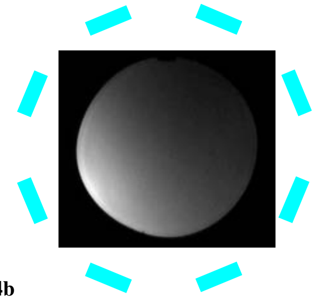


4a



4b

4 Comparison of uncompensated (a) and compensated (b) MR images acquired for a mineral oil phantom. The coil elements are indicated by the blue rectangles. The desired target current distribution was obtained after compensation (Fig.4b).

References: [1] Ibrahim TS,et.al.., Effect of RF coil excitation on field inhomogeneity at ultra high fields, JMRI 19(10):1339-47, 2002 [2] Pauly J, et.al., A k-space analysis of small-tip-angle excitation, JMR.; 81: 43-56., 1989 [3] Katscher U., et.al., Transmit SENSE, MRM 49, 144-150, 2003 [4] Roemer PB, et.al., The NMR phased array, MRM;16(2):192-225., 1990 [5] Hoult DI, et.al., The NMR multi-transmit phased array: a Cartesian feedback approach, JMR., 171(1):64-70., 2004 [6] Jakobus U., IEEE Antennas & Prop. Conf., publ. No. 436, pp 182 – 185, 1997 [7] Graesslin I, et al. [2006] ISMRM 14:129