Feasibility study of a multi-channel pTX-array bodycoil with tray-shielding

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Introduction: In MRI transmit B1 inhomogeneities and SAR limits are well-known issues especially at higher field strength. One approach to overcome those issues is a multi-channel pTx-array. Preferably not only stacked in circumferential [1, 2] but also in the axial direction [3]. Several research groups have already shown their results with 8 elements in one ring. 2 rings behind each other might be a good compromise from the application point of view. Furthermore, one ring could be fed from the front and one ring from the backside of the magnet. But, an extension of the pTx array to more than 2 rings becomes challenging, especially regarding the wiring of the elements in the middle and the decoupling network of the non-adjacent elements. A sufficient decoupling between the individual transmit elements is an essential requirement minimizing cross-talk and protecting the single RFPAs. Due to the missing preamplifier decoupling known from the receive coils a coil design is necessary which intrinsically decouples all elements adequately from each other. Therefore the goal of this study was to build up a 3x8 multi-channel pTx array, which should be scalable to an array with even more elements. In addition to these constraints the requirement of at least -15dB decoupling between each element based on the coil geometry only should also be fulfilled in order to minimize the element cross-talk and the derating of the RFPAs. The following experiment investigates the conventional capacitive decoupling in combination with an RF shielding tray [4] at 3 Tesla.

Methods: Inductive (via overlap) or capacitive (via discrete capacitors) decoupling methods are state of the art to compensate next neighbor coupling. All additional methods to decouple non-adjacent elements would end up in a very complex antenna design. Therefore, our approach for non-adjacent elements was a decoupling via distance and a capacitive decoupling for the next neighbors in axial direction. In circumferential direction we investigated the tray-shield approach.
The following experiment is based on a 67cm RF-screen and a 60cm patient bore. The 3.5cm high array was mounted on the inside of the screen. We built 8 units with tray-shielding of 60cm length into which 3 elements fit. In order to guarantee up to -15dB coupling, the distance of a 10 x 15cm coil to another has to be approximately 30cm. So, one of these units has a 30cm middle element and one additional 15cm element on each side. The middle to one outer element is decoupled via one common conductor tuned to the Larmor condition (capacitive decoupling). To reach a decoupling of 15dB in circumferential direction the single units have to be 10cm wide with a distance of 13cm between each other (see schematic in fig. 1 and the experimental setup in fig.2)
Furthermore, we measured the coupling between each element and the field efficiency per ring by using an 8x8 Butler matrix, we estimated the power requirement for the whole array for a 180° pulse and the necessary breakdown voltages of the components, and determined the impact on the RFPA.

Results: Fig. 3 demonstrates the color-coded decoupling matrix of the full multi-channel pTX-array under no-load condition. All entries are below -14dB. When inserting a load phantom (0.6S/m conductivity) these values move even below -16dB. The worst decoupling is between the middle elements. The outer elements from one unit to the other are decoupled 2dB more than the middle elements. The outer elements within one unit are decoupled more than -18dB. All other non-adjacent elements are decoupled more than -20dB.
Running the middle ring in CP-mode by using an 8x8 Butler matrix, the field in the isocenter is approx. 10dB weaker compared to a conventional birdcage body coil. Thus, the required power for 11.75μT (180° pulse) is 10 times higher. The outer elements show a very poor penetration depth and therefore do not noticeably affect the field in the isocenter. Considering a single loop with and without tray-shielding, a 5dB difference could be observed – related to the geometry mentioned above.

Conclusion: We were able to build a 3x8 multi-channel pTX-array bodycoil with at least -14dB between each element. However, the tray-screen decreases the efficiency of the coil dramatically. Additionally, the low efficiency in comparison to a conventional birdcage design is probably also caused by the low coverage over the entire circumference. As a consequence the RFPA has to be substantially oversized, which results in much higher component load and also in increasing local SAR-effects.
One option to overcome the necessity of such a high decoupling is the use of current-source RFPAs based on low output-impedance amplifier concepts. [3]. Another approach is the usage of circulators to decouple the RFPAs from the feeding ports of the bodycoil.
In summary, we found a setup to overcome the strong coupling between each element of a 3x8 multi-channel pTX-Array. In further investigations, the setup will be improved in order to increase the coil efficiency and to get also an acceptable performance in the receive mode.

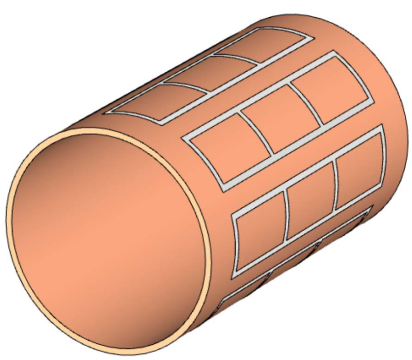

Fig.1 Schematic of 3x8 the pTX BC

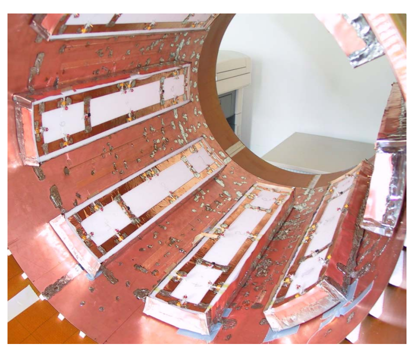

Fig.2 Protoytype of the 3x8 pTX BC

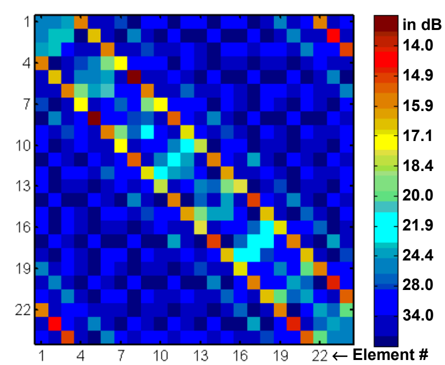

Fig.3 Color-coded decoupling matrix

References:

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[4] Leussler Ch, Shielded Multix Coil Array For Parallel High Field MRI; Patent No.: US 8,013,606 B2; (2011)