Radial Multi-Coil Imaging

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INTRODUCTION: The multi-coil (MC) technique enables the generation of a multitude of magnetic field shapes relevant to MR applications. Magnetic fields are synthesized with the MC approach by a scaled superposition of generic (not necessarily orthogonal) field shapes from individual basis coils and the method stands out by its ability to flexibly trade volume-of-interest, field accuracy and generation efficiency for the experiment at hand [1]. After the application of MC modeling to the homogenization of magnetic fields in the mouse and the human brain [2,3], here we demonstrate the first direct MC imaging application in which magnetic fields are generated by MC modeling and dynamically applied in a regular MR sequence scheme.

METHODS: The applied MR setup was identical to the one used for magnetic field homogenization of the mouse brain at 9.4 Tesla [1]. In essence, a home-built Bolinger antenna was used for RF transmission and signal reception. The MC setup was arranged on a cylindrical surface on the outside of the RF antenna and consisted of 6 rows of 8 coils (diameter 13 mm, 30 turns). Each of the coils was driven individually in the ±1 A range with dedicated amplifier hardware [4]. Spin-echo radial imaging with 96 MC-generated, linear encoding gradients was applied to a structured phantom and reconstructed by filtered back-projection. Multiple images over a 12×12 mm^2 field-of-view were acquired at varying imaging bandwidth to illustrate the trade-off between shape accuracy and achievable field amplitude that is possible with the MC approach. For comparison, reference imaging at identical parameter settings was done with the MR scanner's built-in gradient coils that are based on dedicated wire patterns, one for each gradient term.

RESULTS: Magnetic field modeling with the described MC setup allowed the synthesis of rotating 10 kHz/cm gradient fields over the considered 12 mm field-of-view at (0.061 ± 0.001)% average error (mean ± SD over all 96 encoding steps). An example XY gradient is shown in figure 1 (first row, MC 10 kHz/cm) along with its error distribution (second row). The radial images acquired with these fields (Fig. 2, MC 10 kHz/cm) were artifact-free and virtually identical to those provided by the MR system's built-in gradient coils (Fig. 2, Conventional). Similarly high precision gradient fields could not be generated at higher amplitudes given the severe current limitation of the applied amplifier system. However, the unique flexibility of the MC approach to flexibly trade field accuracy for field efficiency allowed the synthesis of significantly stronger field gradients at moderately reduced generation accuracy with the same setup and amplifier hardware. The (0.12 ± 0.05)% average deviation of the 20 kHz/cm gradient fields stayed close to the noise level and remained invisible in the resultant radial images (Fig. 2, MC 20 kHz/cm) and achieved average errors of (0.59 ± 0.01)% and (0.89 ± 0.01)% for the generation of 30 kHz/cm and 40 kHz/cm gradients, respectively, led only to minor distortions in the image periphery (Fig. 2, MC 30 kHz/cm & MC 40 kHz/cm). For the generation of even stronger, 50 kHz/cm field gradients, amplitude (rather than shape) restrictions of MC fields became the limiting factor of the gradient synthesis. While the generated MC fields still largely resembled the shapes of the radial gradients to be generated, the resultant insufficient amplitude range resulted into a (3.02 ± 0.18)% average error that translated to an effective field-of-view enlargement or, analogously, to a shrinkage of the imaged object within the targeted field-of-view (Fig. 2, MC 50 kHz/cm).

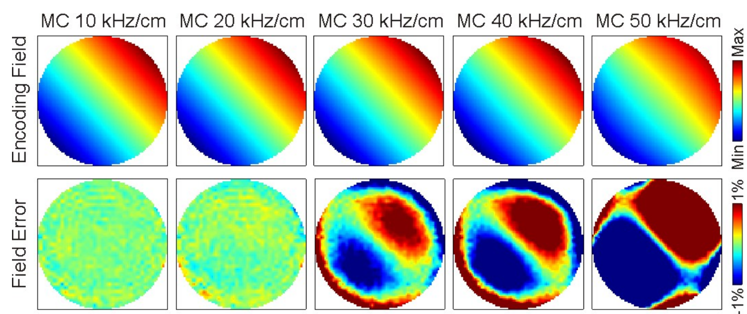

Fig. 1

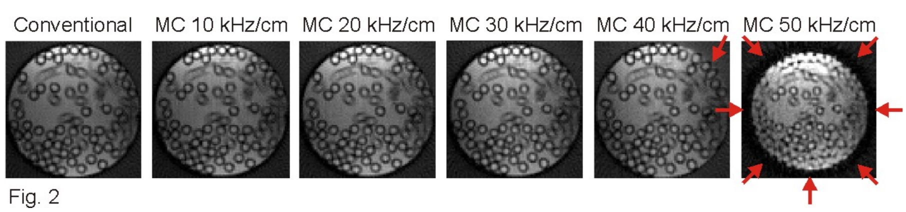

Fig. 2

DISCUSSION: MR imaging based on MC-synthesized, linear magnetic field gradients has been presented. Reasonable imaging bandwidth was achieved with non-optimized MC setup and hardware. The MC approach was furthermore shown to allow multi-fold increases in magnetic field strength and available imaging bandwidth, when perfectly shaped magnetic fields are not strictly required (e.g. for radial imaging). Notably, a similar tradeoff is not possible for the generation of magnetic fields with predefined, dedicated wire patterns. The range of available MC field amplitudes can be further increased with the application of stronger, readily available amplifiers or with the use of more turns per basis coil. Since the MC concept is well understood and experimental magnetic fields are predictable [1], further improvements of MC-based imaging are expected with the application of post-processing methods to consider the known imperfections of the applied magnetic fields for image correction. Identical sequence timing was used in this study for MC and conventional gradient encoding to ensure comparable image contrast, however, available MC switching times as short as 10 µs [4] bear significant potential for faster applications. MC-based imaging is expected to provide an inexpensive alternative for specialized imaging applications for which maximum gradient strength and performance are not required.

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[1] JMR 204:281-289 (2010); [2] MRM 66:893-900 (2011); [3] JMR 212:280-288 (2011); [4] Proc ISMRM 2010, p.1532.