Initial realisation of a multichannel, non-linear PatLoc gradient coil

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Spatial encoding using non-bijective, curvilinear magnetic fields [1] has been introduced to overcome present limitations of gradient performance due to safety limitations from peripheral nerve stimulation (PNS). These PatLoc(parallel acquisition technique using localized gradients) fields allow for higher $\Delta B_{max}/dt$, and therefore faster imaging while staying below the limit of PNS. PatLoc fields may also be adapted to the subjects or organs anatomy, as it is not restricted to rectilinear data acquisition. For practical realisation in cylindrical geometry at least two fields are required, providing two gradients, one in radial and one in circumferential direction to retain some image reconstruction properties. As a continuous circumferential gradient is physically not realizable, a concept of using multipolar fields for spatial encoding was proposed [2]. It is most promising to use this gradient system as a head gradient set. For easy implementation this first functional model is designed for an animal MRI system. This abstract presents the first built gradient coil.

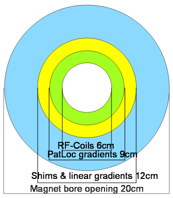

***Fig 1**. Design*

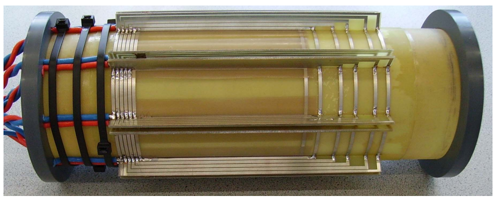

***Fig 2**. First optimized and built coil*

Materials

Based on the design of Cho [3] for a flat surface gradient coil, we chose a head-to-head arrangement of two coils for each coil element. In our initial gradient design the coil consists of eight such coil elements in octagonal arrangement around the z direction. For manufacturing reasons an asymmetrical geometry was chosen rather than the symmetrical layout by Cho. This gradient coil is used as an insert into a 94/20 Bruker BioSpec (Bruker BioSpin GmbH, Germany) with a BGA 12S gradient system. The PatLoc system is placed inside the existing gradient system and replaces the conventional x/y gradients. The z gradient is still be used for slice selection. With RF coils inside, a 6cm opening is left for imaging, see Fig. 1. The position of the transverse wires along the z direction was optimized for a linear B_z in the region of 1-3cm radius over a volume of ±2cm in z direction. The copper conductive paths are 2 mm wide and 35 µm thick, they were etched and treated with tin before being glued on the former with 4,5cm radius. See the final coil in Figure 2.

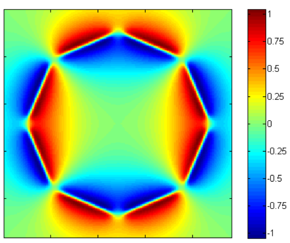

***Fig 3a**. Simulated B_z [10^{-4}T]*

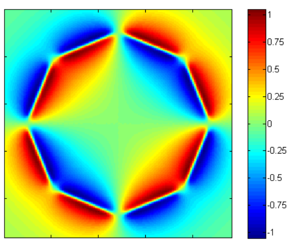

***Fig 3b**. Simulated B_z [10^{-4}T]*

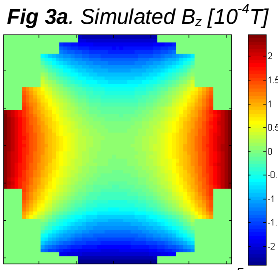

***Fig 4**. Measured B_z [10^{-5}T]*

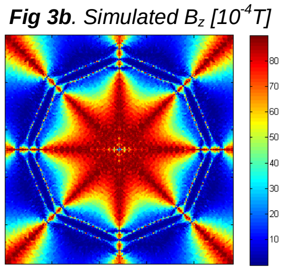

***Fig 5**. Angle between gradients of Fig2 [α]*

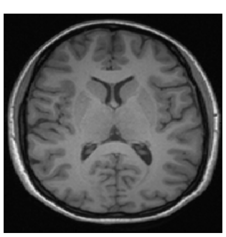

***Fig 6a.** Source image, linear gradients*

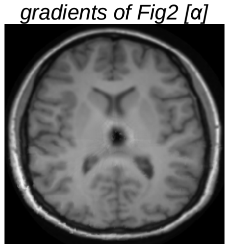

***Fig 6b.** Simulated reco, PatLoc gradients*

Methods

Magnetic field simulations where performed using the Maxwell 3D software package (AnSoft Inc, USA). All simulations assume that the coil elements are driven by 1A and positioned on a radius of 4,5cm. As it is essential to provide two gradients, preferably in plane orthogonal to each other, the fields in Fig. 3 were explored. Only B_z in the x/y plane at the isocenter is displayed. Both fields respectively provide radial gradient as well as circumferential gradient.

The actual field B_z of the gradient coil was measured using a Hall probe. Each coil element was set on 1A current, and the field was measured in 1mm steps only over the interesting region of ±3cm, on the x/y plane at the isocenter, see Fig. 4. Comparison of simulated and measured field shows that both fields are in the same order of magnitude. The measured field, see Fig. 4 was interpolated according to [4], scaled up to a 1024 matrix and rotated by 45° to provide the second field, equivalent to Fig 3b. These fields were used for the simulation of a SENSE-like reconstruction [1, 5].

Results

Combination of both fields provides gradients which are nicely orthogonal in the previously defined region of interest (1-3cm radius), see Fig 5. Therefore these fields allow for orthogonal encoding in local regions where no ambiguities appear. Ambiguities are resolved by acquiring signal with multiple receiver coils and by the SENSE-like reconstruction.

Fig. 6b shows the result of a simulated reconstruction of Fig. 6a using the measured PatLoc field (Fig. 4) and the rotated corresponding one. It can be seen that the gradient in the middle is too small for imaging, so the signal is smeared and reconstruction is impossible (“black hole”) whereas in the periphery the structures of the brain is well resolved.

Discussion & Outlook

If the exact spatial encoding magnetic fields (SEMs) are known, ambiguities and non-linear gradients can be accounted for in the SENSE-like reconstruction. With this approach, higher gradients or higher slew rates can be achieved and overcome present limitations of gradient performance. Scaling these fields/gradients up to 25 cm diameter and with today’s limit of $\Delta B_{max}/dt$ = 20T/s, considering PNS, our gradients can be switched more than 3 times faster compared to a conventional gradient head coil. Other multipolar fields, for example with 6 poles or even more, and appropriate combination of them can also be considered for imaging. Apart from image encoding, multipolar gradients offer interesting opportunities for spatial selection when used as selection gradients. It should also be mentioned that multipolar gradient fields are closely related to the fields of shim coils. Experiments to characterize switching and eddy current behaviour and to use the PatLoc coil for imaging are currently under way.

[1] Hennig et al, Proc.Ann.Meeting ISMRM, Berlin, p.453 (2007); [2] Hennig et al., MAGMA, submitted; [3] Cho et al., J.Magn.Reson 94:471-485(1991); [4] Lee at al. IEEE Trans Vis Comp Graph 3(3):228-244(1997); [5] Schultz et al. submitted to ISMRM 2008

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