PatLoc: Imaging in non-bijective, curvilinear magnetic field gradients

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At the current stage of development, gradient performance for MRI is limited by safety concerns due to peripheral nerve stimulation rather than by technical limitations of gradient coils and/or power supply. Peripheral nerve stimulation is dependant on the maximum local change in magnetic field at the location with the largest field change ΔBr=dB(r)/dt. ΔBr scales with the size of the gradient system for a given gradient field geometry and slew rate dB/dr/dt. Therefore higher slew rate can be achieved with smaller gradient systems. These limitations are a consequence of the inherent unidirectional and linear design of spatial encoding fields leading to the conventional setup of x,y,z-gradients. In order to go beyond these fundamental limits, we present a concept study for spatial encoding by non-unidirectional, non-bijective spatially encoding magnetic fields.

Materials and methods

The basic transformation equation for spatial encoding in Fourier (FT)- imaging is given by

$$I\omega(\omega_{kx}, \omega_{ky}, \omega_{kz}) = FT\left\{\iiint M(x,y,z)\exp\left(-i\gamma\int Bz(x,y,z,t)\,dt\right)dxdydz\right\} \qquad (1)$$

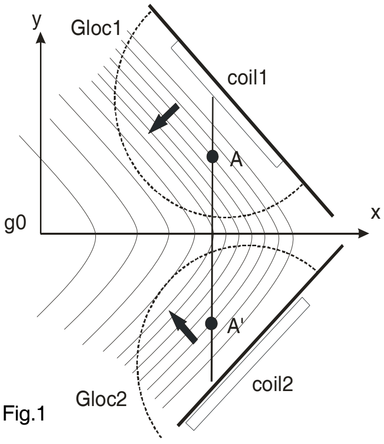

Fig.1

which translates the local magnetization M(x,y,z) into a signal intensity $I_\omega(\omega_{kx}, \omega_{ky}, \omega_{kz})$ in frequency space. Transformation from frequency space to the final image space in spatial coordinates is given by the inverse function of the field change along the spatial coordinates:

$$(x, y, z) = f^{-1}(\omega_{kx}, \omega_{ky}, \omega_{kz}) = 1/\gamma\,(Bz\,(x,y,z)^{-1}) \qquad (2)$$

Using constant, unidirectional and orthogonal gradients the translation from frequency space to Cartesian coordinates is linear with $(r) = (\omega_{kx})/(\gamma * G(r))$ with $r=(x\ y\ z)^{-1}$ and the (spatial) image can be taken directly from the FT of the signal after stepping through the encoding gradients. For arbitrarily shaped spatial encoding fields (SEMs) including non-monotonous fields eq.[1] will become non-bijective leading to ambiguous encoding, i.e. any point in frequency space may correspond to multiple locations in space. This is illustrated in Fig.1 for a curved SEM. Using a conventional x-gradient for encoding in the 2nd direction will produce ambiguous encoding in the regions above and below the x-axis: For any point A there is a mirror point A' with identical spatial encoding. If signal is measured with local coils coil1 and coil2 with different sensitivity profiles, the ambiguities can be resolved by using parallel image reconstruction techniques. The resulting two images in non-linear, but locally unambiguous fields Gloc1 and Gloc2 can then be corrected for distortion.. In general the spatial encoding fields can be subdivided into nc different regions $Gloc_1…Gloc_{nc}$ within which spatial encoding is bijective, such that a minimum of nc different receiver coils with appropriate sensitivity profiles have to be used to produce unambiguous encoding. As a practical implementation for this PatLoc(**par**allel imaging in **loc**al encoding fields)-approach we have simulated radial SEMs as shown in Fig.2A. Field was generated by 8 rectangular current loops in octagonal arrangement. Simulations were performed using an adaptation of a MatLab-routine for numerical solution of the Biot-Savart Law (1). By inversion of the direction of current in alternating loops, a second SEM with alternating tangential local fields is generated (Fig.2B).

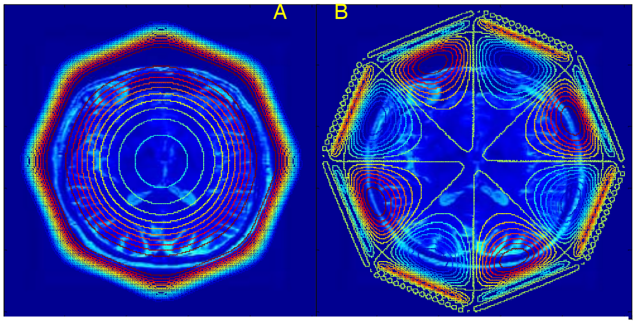

Fig.2 Contour lines of radial (A) and tangential (B) PatLoc-gradients. Superimposed on the head image used for simulations.

Results

Fig.3A shows the image filtered with the sensitivity profile located at the back of the head. Fig.3B shows the distorted image generated from using the polar PatLoc gradients as read- and phase-encoding gradient for a simulated 2DFT-experiment. Backprojection into image space by decoding image 3B with the known inverse of the field profile leads to the 8-fold degenerate image 3C, from which the true image is reconstructed (3D). Comparison of 3D with 3A demonstrates the good image behavior of the ‚polar' PatLoc-gradients. Comparison of the placement of the gradient field with the image location in Fig.2B reveals, that the outer skull is close to the flat top of the gradient, which leads to misalignment and even some mirror artifact at the back of the skull.

Discussion

Imaging with PatLoc –gradients requires exact knowledge of the SEM-profiles in order to account for the drastic distortions generated by non-linear gradient. In addition PatLoc-images inherently will have 'holes' at locations, where the SEMs show flat profiles like in the center of the head in the polar PatLoc gradients shown in our example. On the positive side the slew rate at which the physiological limit is reached, is dramatically higher. For the polar design 4-5 fold higher slew rates can be achieved compared to even optimized conventional head gradient inserts. PatLoc is an extension of the ideas of mulitple region MRI first presented by Oppelt(1) and realized by Parker(2) with the addition degree of freedom afforded by the use of curvilinear encoding fields. This may make the imaging process less intuitive compared to unidirectional x,y,z gradients, but offers significant benefits in terms of the ease of realization as shown by the very simple design used for the polar gradients (Fig.2). The polar gradients shown are just one simple example illustrating the flexibility of the approach. Small organ-dependant PatLoc inserts with dedicated geometries may be developed for different areas of the body and used as inserts just like dedicated RF receiver coils in todays systems.

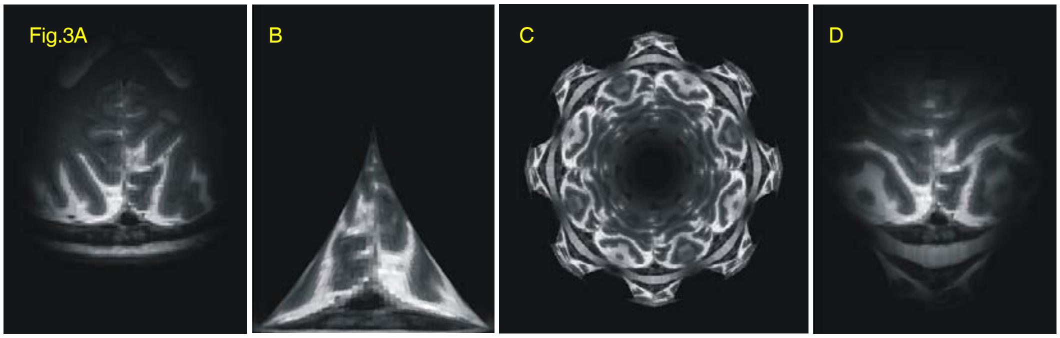

Fig.3A

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