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Receive Coil Arrays and Parallel Imaging for Functional Magnetic Resonance Imaging of the Human Brain

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Abstract—**The use of multiple small receive coils has several advantages over a single larger (e.g. birdcage) coil. With an array of small receive coils, image signal-to-noise ratio (SNR) can be optimized throughout the field-of-view and the image acquisition process can be accelerated by use of parallel imaging (PI) techniques. In an accelerated PI experiment, data are undersampled during acquisition. Subsequently, artifact-free images are reconstructed based on the independently acquired signals from the elements of the receive coil array. PI techniques have recently been applied to functional MRI (fMRI) experiments of the human brain in order to improve the performance of commonly used single-shot techniques like echo-planar imaging (EPI). Potential benefits of PI-fMRI include the reduction of geometrical distortions due to off-resonance signals, the reduction of signal-loss in areas with substantial signal inhomogeneity, increases of the spatial and temporal resolution of the fMRI experiment and reduction of gradient acoustic noise. Although the loss in SNR, inherent to PI, can severely compromise MRI image quality, the effect on fMRI quality, which is governed by the temporal stability of the signal, is often not as severe. On the other hand, PI's potential in mitigating the often severe image artifacts present in single-shot fMRI render it an important tool, in particular with the recent surge in high field MRI applications.**

I. Introduction

Functional MRI puts a number of specific demands on the MRI imaging sequence and hardware. In general, it requires: high temporal resolution ($\geq$ 0.5 image/s); large coverage (whole brain); low artifact level; high temporal stability; and low acoustic noise levels. Furthermore, many functional MRI (fMRI) applications benefit from high field strength, since this increases contrast-to-noise ratio (CNR).

Echo-planar imaging (EPI) and other single-shot techniques are commonly used in fMRI since they fulfill several of the above demands. Most importantly, they allow fast scanning and are more stable than multi-shot techniques. However, EPI comes with drawbacks such as geometrical distortions and T_2*-blurring, as well as high acoustic noise levels, effects which increase with field strength. Several of these drawbacks can be alleviated by combining EPI with accelerated parallel imaging (PI) [1-4].

PI was initially developed to increase image acquisition speed for cardiac MRI applications [5]. Since PI generally results in a decrease in image signal-to-noise ratio (SNR), the benefits for fMRI are not immediately evident. However, most fMRI experiments are primarily limited by physiological noise (temporal instabilities), not image SNR. Under this circumstance, the use of PI will lead to a limited penalty in fMRI sensitivity while allowing reduced artifacts, increased spatial resolution, increased temporal resolution, reduced acoustic noise and/or increased coverage (more acquired slices per unit time). Furthermore, PI appears critical to yield the full potential of fMRI at high field strength ($\geq$ 3 T), where the reduced T_2* significantly affects EPI performance.

PI requires an array of receive coils, the elements of which each have different spatially varying sensitivity profiles. As will be described below, such arrays need to be optimized for PI, and yield additional benefits compared to a conventional coil setup such as increased SNR.

II. Benefits of Receive Coil Arrays for MRI

Over a decade ago Roemer et al. [6] showed that the use of an array of small receive coils has several advantages over a single (e.g. birdcage) coil. Most importantly, SNR can be optimized throughout the field-of-view (FOV) in post-processing. If an object is completely covered with surface coils, the SNR in the center of the object will be equal to that obtained with an identically-sized birdcage coil; however, a higher SNR will be obtained anywhere else in the object, the largest gain being available near the surface of the object.

The concept of parallel imaging was conceived during the same time period [1, 2]. In a PI experiment data are undersampled, however acquisition is performed using multiple autonomous receive coils with spatially (partially) independent sensitivity profiles. This allows computation of the information missing from such undersampled data, lack of which would otherwise lead to image artifacts. The signals from the different coil images are combined in post-processing using knowledge about the spatial sensitivity distribution of the various coil elements throughout the object under investigation, as well as the noise correlation between the different coil elements. This same knowledge is simultaneously applied to improve SNR. SNR and PI-artifact suppression typically increase with the number of coil elements and their mutual independence, and decrease with PI acceleration factor [7]. The PI penalty is spatially variant and commonly expressed by coil geometry factor *g* [4].

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Partly due to limited computing power at the time, both concepts remained relatively obscure until the late 1990ies, when the SMASH [3] and SENSE [4] techniques were proposed. Albeit intrinsically similar, SMASH computes missing *k*-space data whereas in SENSE the fold-over artifacts are removed in image domain.

III. Evaluation of PI Penalty on fMRI Sensitivity

In fMRI, PI can be used to accelerate the experiment. When all other scan parameters remain the same, SENSE MRI with an acceleration factor of R leads to an image SNR reduction of a factor $g\cdot\sqrt{R}$ [4]. This reduction in image SNR does not directly translate into an equivalent reduction in fMRI sensitivity, as will be shown below. Activation levels in blood oxygen-level dependent (BOLD) fMRI experiments are not directly derived from the signal amplitude, but follow from the difference signal between rest and active state. Subsequent division by the level of temporal instabilities (expressed here as the temporal standard deviation (SD) σ_t) of the signal determines the statistical significance of the signal change [8, 9]. Both the intrinsic (image) noise standard deviation σ_i and the physiological noise (described here as the SD of physiological fluctuations, σ_{ph}) contribute to σ_t. Therefore, if image SNR ($\sim\sigma_i^{-1}$) exceeds temporal stability, it can be sacrificed, for example through use of PI, to achieve artifact reduction without substantially affecting the statistical power of fMRI [10]. In [10] it was assumed that σ_i and σ_{ph} are fully independent and Gaussian noise sources and therefore that

$$\sigma_t = \sqrt{{\sigma_{ph}}^2 + {\sigma_i}^2}\ . \qquad (1)$$

In this model, only σ_i is affected by the reduced sampling in accelerated PI, while σ_{ph} remains the same. As was stated above, σ_i increases by $g\cdot\sqrt{R}$ in a rate-R SENSE experiment and therefore

$$\sigma_{t,PI} = \sqrt{{\sigma_{ph}}^2 + {\sigma_{i,PI}}^2} = \sqrt{{\sigma_{ph}}^2 + \left(g\cdot\sqrt{R}\cdot\sigma_{i,noPI}\right)^2}\ , \quad (2)$$

where $\sigma_{i,noPI}$ is the intrinsic SD in a given experiment without SENSE, and $\sigma_{t,PI}$ and $\sigma_{i,PI}$ are the temporal and intrinsic standard deviations in the SENSE experiment with otherwise identical scan parameters [10]. This indicates that the penalty for SENSE use in fMRI solely depends on the relative contribution of physiological noise to the overall temporal SD:

$$\frac{\sigma_{t,PI}}{\sigma_{t,noPI}} = \sqrt{1+\left(g^2\cdot R-1\right)\cdot\left(\frac{\sigma_{i,noPI}}{\sigma_{t,noPI}}\right)^2} \qquad (3)$$

The penalty for PI use in fMRI might therefore not be as severe as would be expected based on decreased image SNR alone. In a σ_t-dominated experiment the measured fMRI activation does indeed suffer the full $g\cdot\sqrt{R}$ penalty ($\sigma_{t,PI}/\sigma_{t,noPI} \approx g\cdot\sqrt{R}$, since $\sigma_{i,noPI}/\sigma_{t,noPI} \approx 1$, see (3)). On the other hand, if σ_{ph} is dominant, application of SENSE will not affect the sensitivity of the fMRI experiment ($\sigma_{t,PI}/\sigma_{t,noPI} \approx 1$, since $\sigma_{i,noPI}/\sigma_{t,noPI} \approx 0$). This model can be used to optimize spatial resolution for a given fMRI experiment.

The above only applies to single-shot PI experiments in which all scan parameters (most notably the echo time TE) remain the same, with exception of the acquisition window duration. In some cases, the shortened acquisition window can be exploited to reduce TE and/or repetition time TR, which would reduce the penalty incurred by PI use. For example, in high-resolution single-shot experiments at elevated field strength, the minimal TE is likely to be above what is generally considered to be optimal for fMRI sensitivity (TE ~ T_2* [11, 12]). In this case, PI could improve sensitivity by allowing the use of a shorter TE. This improvement is in analogy with the improvements with SENSE observed in diffusion-weighted MRI [13]. If a multi-shot imaging sequence is used, and assuming that temporal correlation is negligible, the R-fold reduced sampling can be partly compensated for by close to R-fold improved temporal resolution, so that the resulting fMRI sensitivity penalty is only expected to approach g-factor. Reduced sampling in PI might even reduce σ_t, as it reduces motion-related instabilities [14].

IV. PI-fMRI for Artifact Reduction

A negative side effect of the long readout train used in single-shot techniques like EPI is that the resulting images are prone to geometrical distortions due to off-resonance effects [15]. Also, since a relatively long TE is generally used to attain optimal sensitivity for the BOLD effect, BOLD fMRI data are sensitive to signal loss due to intra-voxel dephasing, which causes signal drops in the images. PI can play a role in reducing both these effects.

Geometrical distortions are a result of the use of a long readout train in the presence of field inhomogeneities in the human head. Such inhomogeneities tend to be most severe in brain regions close to bone-tissue interfaces or air cavities, such as the nasal cavity and in the proximity of the ears, where they can lead to signal loss [16]. Due to such inhomogeneities, spins in some areas of the brain are off-resonance, causing the phase of their signal to change during data acquisition. This phase change is incremented during the acquisition of a train of echoes, as is used in EPI, leading to a linear phase gradient over k-space in the phase encode direction for these spins, which in turn leads to a shift of the signal in the reconstructed image. The size of this effect scales linearly with field strength, thus limiting the applicability of single-shot EPI at high field strength (e.g. 7 T). When using rate-R PI, the length of the readout can be shortened by a factor R, thus reducing geometrical

distortions by a factor of R also. This is the result of the extent of k-space being traversed R times faster, leading to a factor R decrease in the steepness of the aforementioned phase gradient. With otherwise identical scan parameters, this PI-fMRI approach leads to a nominal reduction in intrinsic SNR of a factor $g\cdot\sqrt{R}$ compared to the equivalent non-PI acquisition.

In highly inhomogeneous areas, resulting in a short local T_2*, or at higher field strength where T_2* is globally shortened, the minimum TE of a single-shot EPI experiment might be longer than optimal. Both hardware limitations and the possible induction of peripheral nerve stimulation might render it unfeasible to increase the acquisition bandwidth as a means to reduce acquisition train length. If the resulting phase dispersion is strong this will cause signal loss or even signal dropout. When PI is used for echo-train length shortening the minimal TE will be reduced.

TE reduction is an important advantage for perfusion-based fMRI methods [17, 18]. If the TE is kept short, the amount of signal, and thus SNR, can be increased, as well as its stability. At the same time, the contribution of BOLD signal changes to the experiment can be minimized.

V. PI-fMRI for Increased Spatial Resolution

The spatial resolution of EPI is intrinsically limited by T_2 and/or T_2* signal decay effects during the data acquisition window, which result in blurring [15]. This signal decay causes filtering of the acquired k-space [15], leading to widening of the point spread function. This effect is more severe at higher field strength due to the reduced T_2 and T_2* [13]. When the resolution of a single-shot experiment is limited by T_2 or T_2*, the use of rate-R PI to acquire the data allows acquisition of data with an R-fold increase in nominal spatial resolution (in one dimension) for a given readout train length. Due to the cumulative effect of reduced voxel size and PI-related data reduction the penalty on intrinsic SNR is substantial. Assuming that all other scan parameters remain unchanged, and that intra-voxel inhomogeneities are negligible, the nominal SNR is reduced by a factor of $g\cdot R$.

If signal inhomogeneities are large on a sub-voxel scale they can cause substantial phase dispersion within one voxel, up to the level where it causes significant signal loss. As mentioned above, the TE can be shortened to reduce this effect, but this is not always desirable due to the resulting loss in BOLD sensitivity. Alternatively, the voxel size can be reduced in the direction of the steepest phase gradient.

Both Sodickson et al. [3] and Pruessman et al. [4] briefly refer to the application of PI for spatial resolution increase in their discussion. PI for the reduction of blurring in single-shot imaging was first assessed by Griswold et al. [19], who used SMASH to increase the achievable spatial resolution.

A potential benefit of higher field strength for fMRI, in addition to increased CNR, is an improved specificity to parenchymal activation, partly because of the improved suppression of signal that originates from larger vessels due to the shorter T_2 of deoxygenated blood [20] (but see [21]). Furthermore, the BOLD signal strength has been found to increase at a higher rate than linearly with field strength [22]. Both the increased NMR signal strength at higher field strength and the increased CNR in BOLD-based imaging would allow fMRI at reduced voxel volumes with sensitivity similar to that of conventional experiments at 1.5 T field strength. However, as discussed above, the achievable spatial resolution of single-shot sequences is actually reduced at higher field strength due to reduced T_2 and T_2*. PI should lead to a reduction of the intrinsically achievable voxel size in the phase-encode dimension in EPI by a factor equal to the reduction factor R.

VI. PI-fMRI for Increased Temporal Resolution

For a gradient echo to be optimally sensitive for BOLD signal changes, the TE should be on the order of T_2* [11, 12]. This limits the temporal resolution or coverage increase that can be achieved by application of PI using single-shot sequences like EPI. In most fMRI experiments the TE must be maintained to retain optimal fMRI sensitivity when echo train length is reduced. As a result, the temporal resolution increase, through reduced shot-to-shot TR, can only be a fraction of the reduction in echo train length. Therefore, in the first published application of PI for fMRI, Golay et al. [23] used a PRESTO (Principles of Echo-Shifting with a Train of Observations) sequence [24] to maximize the amount of data acquired per unit time.

VII. PI-fMRI for Acoustic Noise Reduction

The single-shot MRI techniques generally used in fMRI rely on fast gradient switching to encode the signal. This has pushed the development of gradient hardware with higher maximal gradient amplitude and slew rate. The switching of gradients results in changing Lorentz forces on the gradient coil, which lead to high levels of gradient acoustic noise, increasing with magnetic field strength [25]. The typical EPI readout-train results in high levels of scanner acoustic noise in the frequency band for which the human ear is most sensitive, and was found to be the dominant source of acoustic noise [26]. Apart from the discomfort for the patient and even potential lasting damage to the human ear, high sound pressure levels during the exam might affect functional imaging experiments, especially those involving auditory stimulation [27].

It has been demonstrated that PI can be exploited for gradient acoustic noise reduction [28]. This PI-MRI application is based on the reduction in the number of gradient switches that is required to obtain an image of a given spatial resolution. Instead of exploiting this to speed up image acquisition time or resolution, it can be used to substantially reduce the slew rates required to acquire the image. If the spatial resolution and acquisition window duration are unaltered, the ramp times of the readout gradient and the sampling bandwidth can both be reduced by a factor of R for a rate-R SENSE acceleration factor. The factor-R reduced sampling bandwidth leads to a reduction of

a factor R in readout gradient amplitude, therefore reducing the gradient slew rate by a factor R^2. Since the reduction in the number of samples is compensated by the decreased acquisition bandwidth, the intrinsic SNR is not expected to change, except for increases in image noise due to the SENSE g factor.

VIII. Conclusion

Several of the issues that currently hamper fMRI can be addressed by the use of fMRI techniques that employ accelerated parallel imaging. Benefits of PI-fMRI are 1) the reduction of artifacts in single-shot sequences, both geometrical distortions and signal loss due to off-resonance effects, 2) the potential for increases in spatial and/or temporal resolution when employing PI, and 3) the reduction of gradient acoustic noise in order to reduce interaction between the scanner and the fMRI experiment. Although the majority of the PI applications will reduce the intrinsic SNR of MRI, the penalty on fMRI sensitivity can be substantially less.

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