Blipped “yz-Shimming” to Correct for Geometric Distortions in Echo-Planar Imaging of the Human Spinal Cord

Jürgen Finsterbusch[1,2]

[1]Department of Systems Neuroscience, University Medical Center Hamburg-Eppendorf, Hamburg, Germany, [2]Neuroimage Nord, University Medical Centers Hamburg-Kiel-Lübeck, Hamburg-Kiel-Lübeck, Germany

Introduction

Magnetic field inhomogeneities and susceptibility differences cause artifacts in echo-planar imaging like geometric distortions and, in T2*-weighted acquisitions, echo time shifts and signal losses. The latter, roughly, reflect the dephasing effect of the inhomogeneities that is accumulated until the central *k*-space line is covered and can easily be compensated for by an additional pre-dephasing gradient pulse that is applied prior to the echo train (see e.g. Fig. 1a). For the slice direction, this method is known as “z-shimming” and is well-established to correct through-slice dephasing [1-3]. Analogously, compensation gradient pulses can be applied in the phase encoding direction to avoid echo time shifts that modulate the desired image contrast [4]. The geometric distortions are caused by the phase modulation the inhomogeneities induce *during* the echo train and, thus, cannot be addressed by an additional pre-dephasing gradient pulse. Here, an approach involving additional blip gradient pulses in the phase and readout direction (“yz-shimming”) is presented that is able to correct for field-inhomogeneity-induced distortions. Its feasibility is demonstrated in a phantom and the human spinal cord in vivo.

Methods

The basic T2*-weighted echo-planar pulse sequence used is shown in Fig. 1. While a pre-dephasing gradient pulse in the phase-encoding direction (Fig. 1a) is sufficient to compensate echo time shifts and related signal losses, the blips need to be adapted as well (Fig. 1b) to correct also for the geometric distortions that are caused by the phase evolution during the echo train. Analogously, distortions (and signal losses) in the read direction can be considered by introducing appropriate readout blips.

To determine the required blip moments, reference acquisitions with a range of pre-dephasing gradient moments can be performed using the pulse sequence of Fig. 1a. Non-optimal moments do not correct the dephasing effect of the inhomogneities completely which means that the maximum signal intensity in *k*-space is shifted out of the center. This shift yields a linear phase modulation on phase maps after the Fourier transformation, i.e. the optimum compensation setting is characterized by a constant phase in the target region. From this optimum momentum, the (additional) blip momentum can be calculated by taking the echo time and the echo spacing into account, and the pre-dephasing momentum needs to be reduced accordingly.

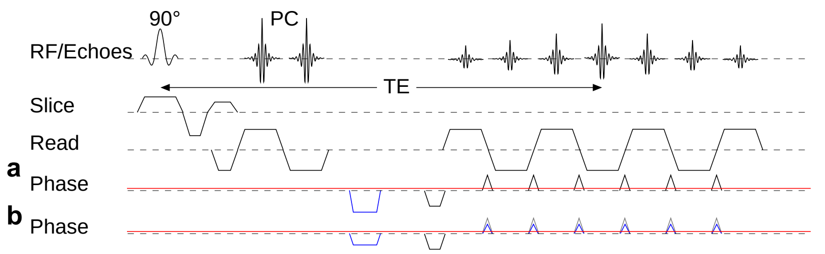

Fig. 1: Basic pulse sequence used in the present study for “y-shimming”. The red lines sketch a background field gradient in the phase-encoding direction that causes an echo time shift and geometric distortions. While the echo time shift can be compensated for (a) by a predephasing gradient pulse (blue), the geometric distortions require in addition (b) a modification of the blips (blue).

Measurements were performed on a 3T whole-body MR system (Siemens TIM Trio) using a 12-channel receive only-head coil. A water phantom and healthy volunteers from which informed consent was obtained prior to the examination, were investigated. T2*-weighted echo-planar imaging was applied with spatial resolutions of $1.0\times1.0\times5mm^3$ (6/8 partial Fourier encoding, FOV 128×128 mm^2) and $2.0\times2.0\times5mm^3$ (FOV 192×256 mm^2) yielding an echo time of 45 ms. In the phantom, magnetic field inhomogeneities were simulated by a shim gradient offset in x and y direction. In the spinal cord, anatomical, T2*-weighted images were acquired with a MEDIC sequence [5] that is very insensitive to in-plane field inhomogeneities and, thus, provides the accurate geometry of the cord.

Results and Discussion

Figure 2 presents results obtained in the water phantom. With a gradient offset in the phase-encoding or readout direction, the phantom appears distorted being either compressed (or stretched for an opposite gradient offset or blip gradient polarity) or sheared. The shift of the maximum intensity of the *k*-space data out of the center yields a linear phase modulation along the direction of the gradient offset, i.e. up-down (phase-encoding direction) and left-right (readout direction), respectively. With an optimum compensation moment and blipped yx-shimming, the shape and size of the phantom is retained and the phase map shows an almost homogeneous phase with a pattern very similar to that without field inhomogeneities. The spinal cord images (Fig. 3) demonstrate the feasibility for in vivo application. With a blipped yx-shimming the geometry of the spinal cord seems to be more reliably depicted

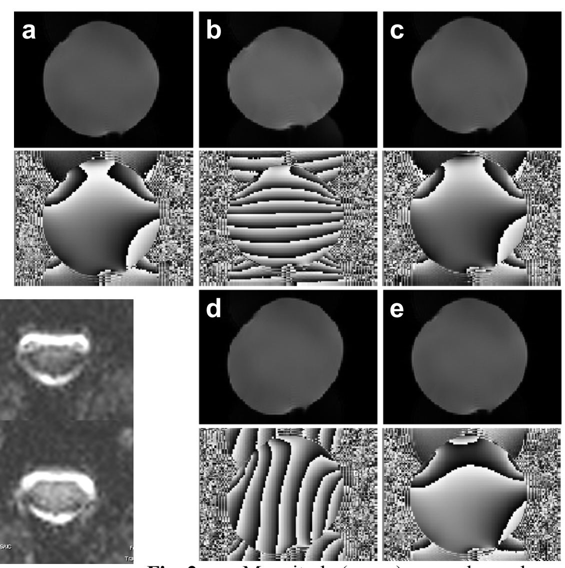

Fig. 2: Magnitude (upper) and phase images (lower) of a phantom for (a) optimized shim and (b-e) simulated field gradients in (b,c) phase encoding and (d,e) readout direction. (b,d) Without compensation, the phantom appears distorted, either (b) compressed or (d) sheared. (c,e) With appropriate additional pre-dephasing and blipped gradient pulses the distortions can be corrected

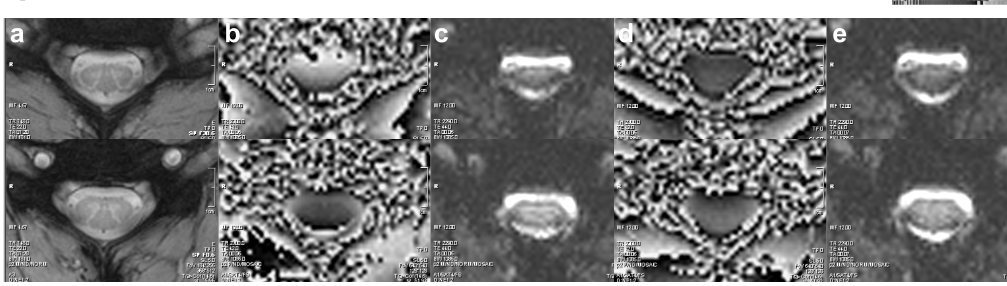

Fig. 3: Examples for two slices within the human spinal cord in vivo: (a) anatomical reference, standard (b) phase and (c) magnitude images and (d) yx-shimmed phase and (e) magnitude images. The latter show a more homogenous phase in the cord and seems to better depict the anatomy

References

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