Requirements on the Accuracy of Navigators for Prospective Motion Correction in High Resolution MR Imaging

M. Zaitsev[1], O. Speck[2], and J. Hennig[1]

[1]Dept. of Radiology, Medical Physics, University Hospital Freiburg, Freiburg, Germany, [2]Dept. Biomedical Magnetic Resonance, University of Magdeburg, Magdeburg

Introduction

Prospectively navigated MR imaging is becoming increasingly popular to overcome present measurement time and/or resolution limitations in such varying application fields as liver, cardiovascular, joint or head imaging. Moreover, navigated imaging appears to be the only way to overcome the limitation for isotropic resolution in the human head of about 500µm. Indeed, to achieve reasonable signal-to-noise ratios (SNR), measurement times of about 10 to 30 minutes are required, in which involuntarily movements of the order of millimetres are unavoidable [1]. Despite of the popularity of navigators, to our knowledge no data on the required navigator accuracy are available in literature. Here, formalism is developed to analyse statistically the image artefacts introduced by the prospective motion correction of each single k-space line. The results enable one to formulate the requirements on the navigator accuracy based on the imaged sample properties and desired resolution.

Theory

K-space signal intensity in traditional spin warp imaging, ignoring saturation, relaxation and dephasing effects can be given as: $S(\mathbf{k})=\int\rho(\mathbf{r})e^{i\mathbf{kr}}d\mathbf{r}$, where $\rho(\mathbf{r})$ is the sample signal density. Prospective motion correction in absence of motion adds uncertainty into the imaging place position relative to the sample, that is:

$$\tilde{S}(\mathbf{k})=\int\rho(\mathbf{r})e^{i\mathbf{k}(\mathbf{r}+\Delta)}d\mathbf{r}=S(\mathbf{k})e^{i\mathbf{k}\Delta}\approx S(\mathbf{k})+\eta(\mathbf{k}). \qquad (1)$$

Thus, small noise in the position introduces an additive k-space noise $\eta(\mathbf{k})=i\mathbf{k}\Delta S(\mathbf{k})$. Fourier transform of Eq. (1) yields

$$I(\mathbf{r})=\int\tilde{S}(\mathbf{k})e^{-i\mathbf{kr}}d\mathbf{k}=\rho(\mathbf{r})+\eta(\mathbf{r}). \qquad (2)$$

Considering, that in gradient echo imaging with line-by line correction the position noise depends solely on the phase encoding step, that is $\mathbf{\Delta}=\mathbf{\Delta}(k_y)$, the noise-like term in Eq. (2) can be expressed as follows:

$$\begin{aligned}\eta(\mathbf{r})&=i\int\mathbf{k\Delta}(k_y)S(\mathbf{k})e^{-i\mathbf{kr}}d\mathbf{k}\\&=i\iint k_x\Delta_x(k_y)S(k_x,k_y)e^{-ik_xx}e^{-ik_yy}dk_xdk_y+\\&+i\iint k_y\Delta_y(k_y)S(k_x,k_y)e^{-ik_xx}e^{-ik_yy}dk_xdk_y\\&=\eta_x(\mathbf{r})+\eta_y(\mathbf{r}),\end{aligned} \qquad (3)$$

where $\eta_x(\mathbf{r})$ and $\eta_y(\mathbf{r})$ are the image noise components originating from x any y components of the position noise, respectively. The noise-like components in the above equation can be further analysed statistically in terms of their variance. Assuming the white uncorrelated noise with $\langle\Delta x(k_y)\Delta x(k_y')\rangle=\sigma^2\delta(k_y-k_y')$ and averaging $\eta_x^2(\mathbf{r})$ and $\eta_y^2(\mathbf{r})$ over the realisations of the position noise it is possible to show that

$$\begin{aligned}\langle\eta_x^2(x,y)\rangle&=\sigma_x^2\frac{N_y}{FOV_y}\left\langle\left(\tfrac{\partial}{\partial x}\rho(x,y)\right)^2\right\rangle_y\\\langle\eta_y^2(x,y)\rangle&=\sigma_y^2\frac{N_y}{FOV_y}\left\langle\left(\tfrac{\partial}{\partial y}\rho(x,y)\right)^2\right\rangle_y,\end{aligned} \qquad (4)$$

where N_y is the image matrix size in phase encoding direction. The total image noise variance equals the sum of the above components, $\langle\eta^2(x,y)\rangle=\langle\eta_x^2(x,y)\rangle+\langle\eta_y^2(x,y)\rangle$.

Methods

To verify the theoretical conclusions experiments were carried out on a 3T Magnetom TRIO system (Siemens Medical Solutions, Erlangen, Germany) using a single channel birdcage coil. Standard gradient echo sequence was modified to enable external real-time motion position input. Imaging was done in a stationary phantom with position noise simulated using a random number generator with a white Gaussian distribution [2]. Magnitude and phase images were reconstructed on the scanner to enable complex valued data analysis.

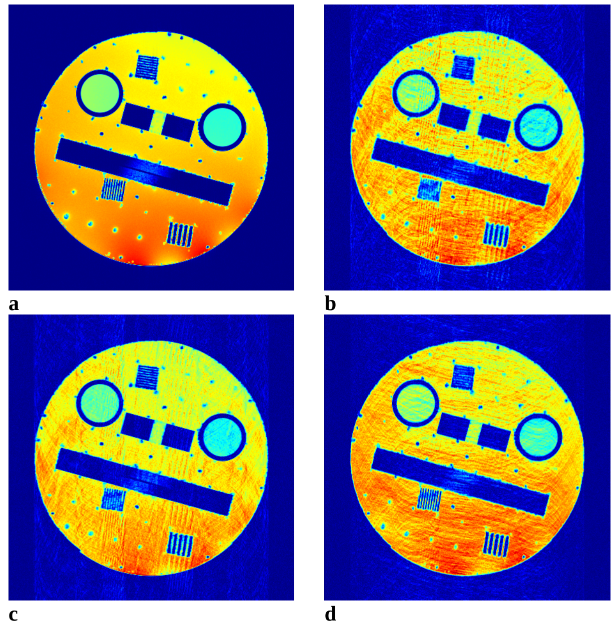

Fig. 1. Phantom images acquired (a) without navigator noise, (b) with isotropic noise of 0.5mm, (c) with the same noise in x and (d) y directions. It is apparent, that mostly vertical edges generate artifacts in figure (c) and horizontal ones in figure (d)

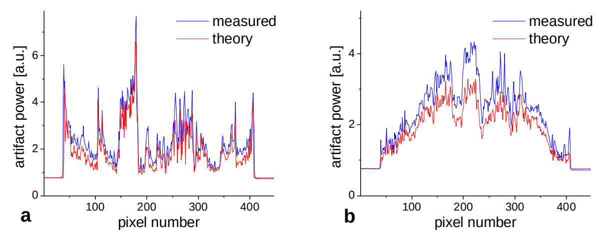

Fig. 2. Mean artifact power generated by the x component of the navigator noise (a) and y (b). Blue line is the measured value, red one is based on the theoretical prediction from Eq. (4).

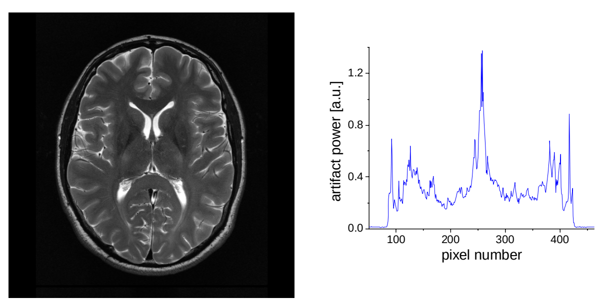

Fig. 3. High resolution brain image (left) used as template to calculate the predicted in-vivo artifact power (right).

Results and Discussion

In Fig. 1 sample results of the phantom imaging experiment with 0.5x0.5mm in-plane spatial resolution and simulated navigator noise with σ=0.5mm are presented. The mean artefact power calculated based on the experimental results and predicted by the theory based on Eq. (4) and phantom geometry for σ=0.5mm is presented in Fig. 2. In agreement with a theoretical prediction the artefact power was found to scale linearly with the matrix size and inversely with FOV (or pixel size). In the regime of navigator noise lying under the pixel size the artefact power scales quadratically with the variance of the navigator position. In Fig 3 human brain image is presented along with the calculated artefact power distribution. The global mean normalised artefact power of this image equals 5.2e-3. Knowledge of this number allows to predict the effective SNR due to the artefacts, e.g for the image matrix of 256x256 with in-plane resolution of 1mm and navigator accuracy of 1mm the expected eSNR=10%. For the practical applications it is desired for eSNR to be notably smaller than native image SNR. From the above example it becomes clear, that in typical brain imaging applications navigator noise shall be a factor of 5 to 10 less then the pixel size.

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