Sensitivity to RF and B_0 Field Imperfections in Continuous Moving-Table MR Imaging

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Introduction

In continuous moving-table MRI with coronal or sagittal plane orientations [1-3], each line of *k*-space is affected in a somewhat different way by the unavoidable system imperfections relating to the RF, B_0 and gradient fields. Thus, phase errors that decrease the image quality only marginally in static or multi-station imaging may lead to considerable artefacts in the case of continuous motion. When *k*-space is traversed in a linear fashion, the artefacts are still moderate and are observed as stripe-like patterns occurring at regular intervals spaced one FOV apart. However, when *k*-space is scanned in a segmented, interleaved fashion, which is studied here, the artefacts may become much more serious because adjacent lines of *k*-space are acquired at distant locations within the FOV. Such segmented strategies are of great interest in view of contrast enhanced techniques where preparation pulses are interleaved in the *k*-space traversal.

Theory

For simplicity, but without loss of generality, the 2D imaging case in a coronal (y,z) plane is considered here, where z denotes the direction of motion. It is assumed that z is also the readout direction. As major sources of image artefacts, phase errors caused by non-uniform RF excitation and reception, φ_{rf}, and spatially varying off-resonance, $\Delta\omega$, are considered here; for gradient non-linearity, see e.g. [4]. Let A denote an amplitude combined from both the RF send and receive coil sensitivities, Δk the increment in *k*-space and t_m the time elapsed since RF excitation for sample m. Then the MR signal, S_{nm}, at phase-encode step n and readout sample m for a voxel located at y,z can be written as (steady state, relaxation neglected)

$$S_{nm} = A(y,z) \exp i2\pi \left[n\Delta k_y y + m\Delta k_z z + \varphi_{rf}(y,z) + \Delta\omega(y,z)\, t_m \right] \qquad (1)$$

Under motion, the location of a voxel is mainly a function of the time τ_n when phase-encoding n is acquired since the motion during the sampling period is small compared with the voxel size and can be neglected, i.e. $z = z_0 + v\tau_n$, where v denotes the table velocity. Thus, Eq. (1) becomes

$$S_{nm} = A(y, z_0 + v\tau_n) \exp i2\pi \left[n\Delta k_y y + m\Delta k_z z_0 + m\Delta k_z v\tau_n + \varphi_{rf}(y, z_0 + v\tau_n) + \Delta\omega(y, z_0 + v\tau_n)\, t_m \right] \qquad (2)$$

The first two terms in the exponent describe the ideal signal. The third term is independent of the spatial coordinates so that it can be readily eliminated by demodulation (or shift after FT), which compensates the table motion. The MR signal after motion compensation may thus be written

$$S_{nm} = S0_{nm}\, A(y, z_0 + v\tau_n) \exp i2\pi \left[\varphi_{rf}(y, z_0 + v\tau_n) + \Delta\omega(y, z_0 + v\tau_n)\, t_m \right] \qquad (3)$$

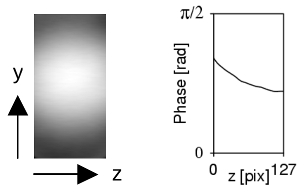

Fig. 1. Receive coil sensitivity, amplitude (left) and phase along magnet axis (right).

where $S0_{nm}$ denotes the ideal signal. The variation of the amplitude A gives only a very small image error in practice (convolution with a narrow function). But the phase terms may be seriously detrimental to image quality, depending on how τ_n varies with *k*-space line n. It can be shown that time t_m may be replaced, as a good approximation, by a constant value, namely, half the acquisition window. Phase errors from both the non-ideal RF and B_0 fields then have the same form. A point to note is that $\Delta\omega$ in the above equations does not include the susceptibility-induced contribution to the inhomogeneity that moves along with the voxel. This contribution is omitted here, for simplicity, because it is the same for all phase-encoding steps and does not lead to artifacts specific to the table motion.

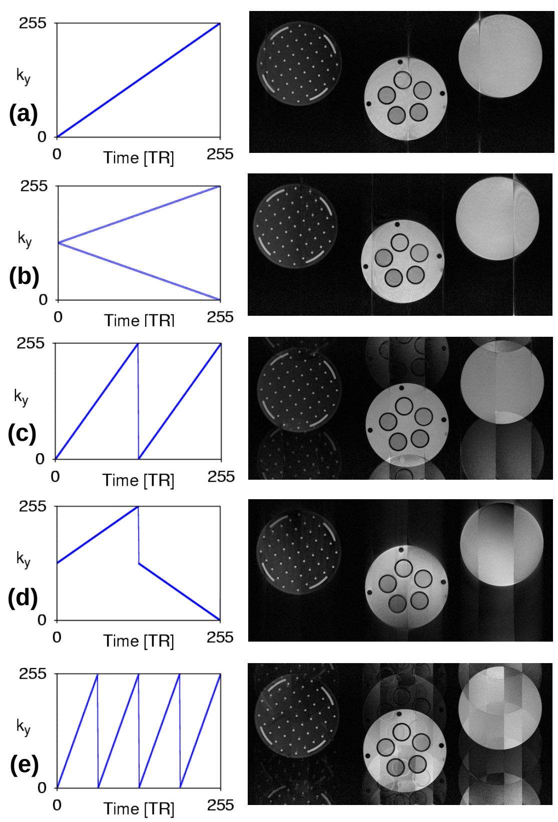

Fig. 2. Modulus images for different acquisition schemes.

Method

Experiments were performed on a 1.5 T whole-body scanner (Gyroscan Intera, Philips Medical Systems). A 2D spoiled gradient echo sequence with TR/TE = 18.5/4.2 ms and 15 degrees flip angle was used with different orders in which *k*-space was traversed. A FOV of 170 mm in z (motion) direction and 340 mm in y direction was chosen with a matrix of 128 x 256. The relatively small FOV dimensions were chosen to focus on phase errors due to RF-coil properties. The body coil was used for excitation and a surface coil for signal reception. A stepping motor was used to pull the patient table with v = 36 mm/s. Image reconstruction was performed as described in [2], except that the amplitude variation of the RF sensitivity was corrected for each line in *k*-space. Figure 1 sketches the properties of the receive coil used. The phase varies approximately 0.3 rad over the imaging FOV along the direction of motion.

Results

A selection of results is shown in Fig. 2. The linear and high-low acquisition schemes (Fig. 2a,b) are relatively robust and show only the typical stripe artefacts. Serious degradation of image quality, seen as ghosting, occurs in the segmented acquisitions (Fig. 2c,d,e). The explanation lies in the increasingly adverse effect of the phase errors on the image quality as neighboring lines of *k*-space are acquired at more distant locations from each other. The effect is mainly attributed to phase variations of the receive-coil sensitivity (confirmed by simulations), with some minor contributions from the send coil and B_0 field inhomogeneity. Effects from a disturbance of the steady state can be excluded since no preparation pulses were applied. Also, eddy currents were negligible here since corresponding static images showed no artefacts.

Conclusion

Segmented *k*-space acquisitions in continuous moving-table MR imaging are very sensitive to RF-coil phase variations and B_0 inhomogeneity, and the same is to be expected for gradient non-linearity at the borders of the FOV. To make such acquisition methods useful in practice, optimized hardware and adequate software corrections are required. Moving-table methods in which images are acquired in a thin transverse plane (see e.g. [5]) do not suffer from these types of artefacts but have the disadvantage of lower SNR because of their 2D orientation.

References

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