# Updating of MRI Gradients Using a Infrared Tracking System to Compensate Motion Artifacts

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## Introduction

Functional MRI (fMRI) is a non-invasive imaging technique that is used to investigate cerebral function. Patient motion remains a significant problem in many MRI applications, including fMRI, cardiac and abdominal imaging, and conventional long TR acquisitions. Many techniques are available to reduce or to compensate for bulk motion effects, such as physiological gating, phase-encode reordering, fiducial markers, fast acquisitions, image volume registration, or alternative data acquisitions strategies such as projection reconstruction, spiral and PROPELLER. Navigator echoes are used to measure motion with one or more degrees of freedom; the motion is then compensated for either retrospectively or prospectively. An orbital navigator (ONAV) echo captures data in a circle in some plane of *k*-space, centered at the origin [1]. This data can be used to detect rotational and translational motion in this plane, and to correct for this motion. However, multiple orthogonal ONAVs are required for general 3D motion determination, and the accuracy of a given ONAV is adversely affected by motion out of its plane. Methods capable of correcting for head motion in all six degrees of freedom have been proposed for human positron emission tomography (PET) brain imaging [2]. These methods rely on the accurate measurement of head motion in relation to the reconstruction coordinate frame.

Implementing a similar technique in MRI presents additional challenges. Foremost the tracking system and the MRI system have to be compatible. High magnetic fields >= 1.5 Tesla in magnetic resonance imaging systems require that the tracking camera system be positioned a sufficient distance from the MRI system to ensure proper function and safety. Functional MRI also proves challenging because of the high spatial accuracy (RMS <0,3mm) required by the complete measurement chain with a small latency time (<30ms) of the tracking system. The coordinate system of a MRI scanner is controlled using magnetic field gradients and frequencies in the sequence. Determining a precise relationship between the spatial varying magnetic field gradients and the spatial tracking information is necessary to compensate for motion artifacts.

## Materials and Method

The tracking system reports the position and orientation of rigid targets fitted with "passive" retro reflective markers in six degrees of freedom (DOF) using two progressive scan cameras synchronized by a frame grabber card in a standard PC [3,4]. Several targets can be tracked simultaneously at a sampling rate of up to 25 Hz, with a quoted positional accuracy less than 0.3 mm (RMS). Communication with the MRI host computer takes place over a TCP/IP connection. Passive targets consist of at least three coplanar reflective markers. All retro reflective markers are filled with doped water to be detectable for both MR and the tracking system. The point of origin of the tracking system is transformed to the physical center of the gradient system to overlap both coordinate systems. The tracking system returns, in the overlapped coordinate frame, target orientation as a unit quaternion and target position as 3 dimensional vector (6 DOF). Every translation and rotation value is stored in a log file. The rotation is measured by rotating a phantom 10 times by one degree. MR images are acquired with and without motion correction. The motion correction is performed prospectively by updating the slice positions and orientations based on the tracking data.

## Results and Discussion

The determined positional accuracy was less than 0.1 mm (RMS). The mean value of the measured reproducibility of the rotation and translation was 0,008/0,067. No artifacts were detectable in the MR images originating from interactions of the MR system and the tracking system. The latency of the whole measurement chain including the reconstruction PC was less than 25ms. It is possible to optimize this latency by supporting the gradient adaptation calculation directly in the acquisition system rather than in the image reconstruction as in the current implementation.

Incorporation of the real-time 3D prospective motion correction demonstrated here in fMRI studies will allow to use all data from non-cooperative subjects even during periods of fast head motion within the acquisition of one volume.

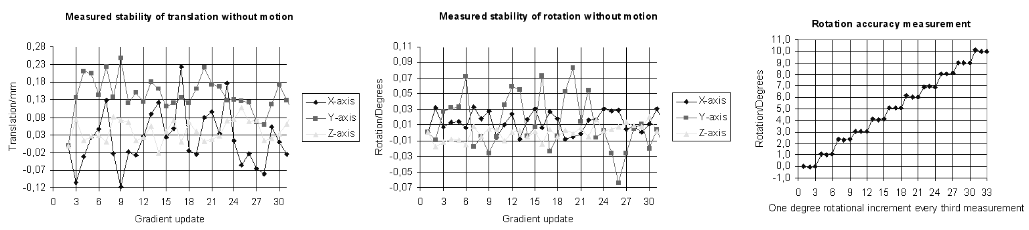


Figure 1: The peaks of the translation and rotation values during different positions of the phantom shown left and middle. The right figure presents the one degree rotation steps ten times. The time to acquire these data took twelve minutes.

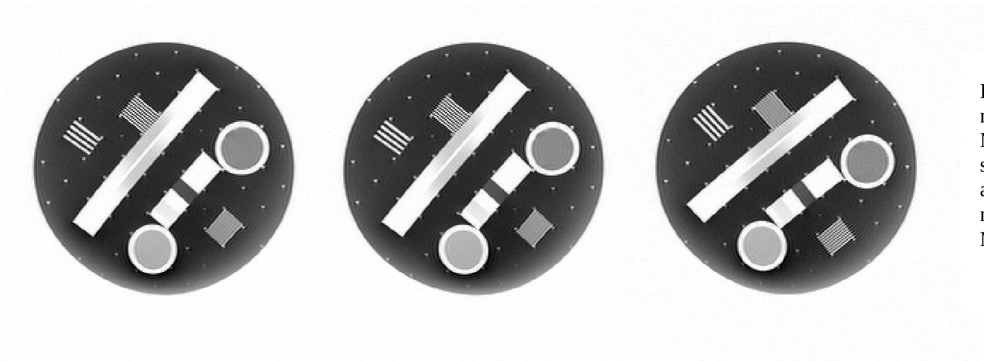

Figure 2: Left the original and middle the motion compensated image by updating the MRI gradients with the data of the tracking system. On the right side no motion correction is activated in the MR scanner. The 10 degree motion is clearly visible. The negatives of the MR images are used.

## References

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