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Joint reconstruction of non-overlapping magnetic particle imaging focus-field data

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Abstract
The focus field is a key component to enable clinical applications in magnetic particle imaging (MPI). Due to physiological constraints, the method of choice is to place the focus of a small acquisition volume at various static positions in space and acquire the full field-of-view using a multi-station approach. In the first experiments, overlapping drive-field patches were used and boundary artifacts between drive-field patches were reduced using image processing. In this work we show that artifact-free reconstruction of non-overlapping focus-field data is feasible by using a joint reconstruction algorithm. This enables maximum scanning efficiency in multi-station focus-field experiments, which is key for reaching sufficiently short acquisition times to image the human heart.

Keywords: magnetic particle imaging, focus field, image reconstruction, scan efficiency

(Some figures may appear in colour only in the online journal)

1. Introduction

Magnetic particle imaging (MPI) is a non-invasive method for determining the spatial distribution of super-paramagnetic nanoparticles—so-called SPIOs—in-vivo. The major milestones during the development of MPI were the invention by Gleich et al. (2001), the first scientific publication by Gleich and Weizenecker (2005), the first in-vivo mouse measurements by Weizenecker et al. (2009), and the introduction of the focus field by Gleich et al. (2010).
In the most basic MPI experiment a static gradient field with a zero crossing at the center of the scanner bore is used. By applying homogeneous drive fields in all three spatial directions the so-called field-free point (FFP) is moved through the measuring field while the change in the particle magnetization is recorded using receive coils. The area that is covered by the FFP trajectory is named the drive-field field-of-view (FOV). It is determined by the ratio between the drive-field amplitude and the gradient strength of the FFP field. Due to physiological constraints (e.g. specific absorbing rate (SAR) and peripheral nerve stimulation (PNS) (Saritas et al 2013, Schmale et al 2013)), the amplitude of the drive field is bounded. Considering a drive-field strength of 10 mT, a gradient strength of 2 Tm⁻¹ in z direction, and −1 Tm⁻¹ in x and y directions, the drive-field FOV is of size 2 × 2 × 1 cm³ and thus too small for human applications.

In order to mitigate the limitation of the drive-field amplitude and in turn the size of the drive-field FOV, Gleich et al (2010) introduced the focus-field concept (also named partial FOV scanning by Goodwill and Conolly (2011)). The homogeneous focus field is superimposed onto the FFP gradient field and leads to a shift of the FFP in space. The low focus-field frequency of a few Hz allows to move the drive-field FOV slowly in space. Due to the low frequency, much higher amplitudes can be applied and in turn much larger volumes can be imaged.

In a static focus-field approach, one first images a volume with the drive field and then moves the center of the drive-field FOV to the next position to acquire the next volume. In this way, the full FOV is sequentially imaged using a multi-patch acquisition scheme as illustrated in figure 1. The volume captured at a certain focus-field position is also named patch (Gleich et al 2010). In this paper only a static focus-field approach will be considered. First results of dynamic focus-field acquisition were published by Rahmer et al (2012).

While the instrumentation of MPI focus-field scanners has been shown to be feasible (Gleich et al 2010), little effort has been devoted to optimizing focus-field sequences and reconstruction algorithms. One challenge is that even particles outside of a drive-field patch are weakly excited by the drive field and thus contribute to the signal generation. Rahmer et al (2011) and Grüttner et al (2012) have found that boundary artifacts disturb the image when reconstructing drive-field patches separately. To cope with this issue, overlapping patches were used and to reduce boundary artifacts a continuation algorithm was applied in a post-processing step. The focus-field overlap leads to a lower scanning efficiency as several parts of the full FOV are imaged twice. The overlap of the 3D volume acquisition, reported by Rahmer et al (2011), was 56.4%. Note that non-overlapping multi-patch acquisition was recently also discussed for microscopic cell imaging by Bise and Sato (2015).

2. Methods

The purpose of this work is to show that focus-field acquisition and artifact-free reconstruction of non-overlapping focus-field data is feasible by using a joint reconstruction algorithm.

In order to avoid a model/measurement mismatch we use the calibration-based system-matrix approach where the full system matrix is determined by shifting a small delta sample through the FOV while measuring the system response, (Knopp and Buzug 2012). In contrast to Rahmer et al (2011) we acquire a dedicated system matrix $S_l$, $l = 1, \ldots, L$ for each of the $L$ focus-field stations. The corresponding measurement vectors at different focus-field stations are named $u_l$. One further difference to the approach taken by Rahmer et al (2011) is that we measure the system matrices $S_l$ not only within the corresponding drive-field patch but in the
full FOV. This is illustrated in figure 2, where several system matrix rows are shown for four non-overlapping focus-field stations.

Let $\mathbf{c}$ be the vector that contains the discretized particle concentration in the full FOV. Then each focus-field measurement forms a linear system of equations $S_l \mathbf{c} = u_l$. When solving these systems independently, the reconstructed concentrations $c_l$ will only be partly equal to $\mathbf{c}$ as no spatial encoding happens at FOV parts that are not covered by the drive field.

To circumvent handling of boundary artifacts we propose in this work to solve the joint linear system of equations

$$
\begin{bmatrix}
S_1 \\
\vdots \\
S_l \\
\end{bmatrix}
\mathbf{c} =
\begin{bmatrix}
u_1 \\
\vdots \\
u_l \\
\end{bmatrix},
$$

In this way, all boundary handling is inherently included in the linear system to be solved. For solving the system (1) we use the regularized Kaczmarz algorithm that is known to rapidly converge for MPI system matrices (Knopp et al 2010). The regularization parameter is manually chosen based on visual inspection of the reconstructed images.

Experiments were performed with a Philips/Bruker preclinical MPI scanner. For simplicity, we restrict our study to 2D experiments that were carried out using the $xy$ plane of the scanner (coronal plane). The gradient strength was $1.25 \text{Tm}^{-1}\mu_0^{-1}$ in both directions while the drive-field amplitude was chosen to be $10 \text{mT}\mu_0^{-1}$. This results in drive-field patches of size $16 \times 16 \text{mm}^2$. The drive-field frequencies are given by $f_x = 2.5 \text{MHz}/102 = 24509.8 \text{kHz}$ and $f_y = 2.5 \text{MHz}/96 = 26041.7 \text{kHz}$ leading to a drive-field repetition time of $T_R = 0.6528 \text{ms}$. The resulting Lissajous sampling trajectory has a commensurable frequency ratio of 17/16.

We acquired four drive-field patches with the center shifted to $\pm 8 \text{ mm}$ in both directions. Hence, the patches have no overlap and cover a total area of $32 \times 32 \text{ mm}^2$. Due to current restrictions of the scanner software we had to acquire the drive-field patches in separate scans, leading to pause times of about 5 s. This is, however, no hardware limitation and will be solved
in a future software version of the scanner. The system matrices were acquired on a grid of size $32 \times 32$ covering an area of $40 \times 40$ mm$^2$ using a delta sample of size $23$ mm$^3$.

We used three different particle phantoms that cover all focus-field stations. Resovist is used as tracer for setting up the phantoms. Phantom (a) consists of five dots that are positioned like the dots on a shuffle. Each dot consists of 20 $\mu$l Resovist filled into a PCR tube. Phantom (b) is made by using a flexible tube with an inner diameter of 1.3 mm filled with Resovist representing the letter U. Phantom (c) consists of $5 \times 3$ cylinders with 3 mm diameter and 4 mm length. The gap between the boundaries of the cylinders is also 3 mm. While phantom (a) and (b) are filled with undiluted Resovist (0.5 mol(Fe) l$^{-1}$), phantom (c) contains a Resovist concentration of 0.05 mol(Fe) l$^{-1}$. Pictures of all three phantoms are shown in figure 3.
3. Results

The reconstructed MPI focus-field data are shown in figure 4. The results for the separate reconstruction of the focus-field stations are shown in the first four columns, while the last column shows the results for the joint reconstruction of all stations. One can clearly see that the separate reconstruction results have boundary artifacts in regions that lie outside the drive-field patch. In contrast, the joint reconstruction does not suffer from artifacts at the boundary between drive-field patches for all three phantoms.

Note that the regularization parameter has to be chosen higher if the drive-field patches are reconstructed separately in order to keep the boundary artifacts low. The joint reconstruction results thus appear to be slightly sharper.

4. Discussion

In the present paper we have shown that artifact-free reconstruction of non-overlapping focus-field MPI data is feasible by using a joint reconstruction algorithm. Our approach was to minimize all possible sources leading to an inconsistent or incomplete reconstruction problem. First of all, a dedicated system matrix has been acquired for each focus-field station although the matrices of different stations would theoretically be equal except for a shift in space. In practice they differ due to inhomogeneities of the applied magnetic fields. Second, the system matrix measurement was not restricted to the region scanned by the drive field. Instead, the system matrices were acquired in the full FOV for all focus-field stations. The importance of this improvement can be understood by looking at the system matrix rows shown in figure 2. For frequency component $k = 34$ the signal smears over the full FOV and is not restricted to the region scanned by the FFP trajectory. Restricting the matrix measurement to the scanned drive-field region would lead to an incomplete system at the boundaries. Furthermore, the overlapping system matrix structure motivates the formulation of a joint linear system for all drive-field patches because it takes the coupling between focus-field stations into account.

For the presented 2D measurements and the small image sizes our approach was computationally feasible. The approach will reach its limits if 3D measurements, more focus-field stations, and larger image sizes are considered. The first issue is that the dedicated acquisition of individual focus-field system matrices is too time-consuming for large image sizes given that a single measurement lasts 6 h even for small image sizes of $34 \times 20 \times 28$ (Weizenecker et al 2009). Therefore, we plan to use the compressed sensing approach that was introduced by Knopp and Weber (2013) and allows to significantly shorten the calibration time. The second issue is that for larger image sizes, the size of the joint system matrix reaches the main memory limit of the reconstruction computer. Therefore, we plan to apply the matrix compression

![Figure 3. Particle phantoms used to validate the joint focus-field reconstruction.](image-url)
technique that was proposed by Lampe et al (2012). It allows to keep the joint system matrix into the computer’s main memory for rapidly solving the joint imaging equation.

With the proposed joint reconstruction approach it is not necessary to scan the drive-field patches with overlap. Hence, the scanning efficiency is considerably increased which will be essential in time critical applications such as imaging of the human heart. Considering a gradient strength of 1 $\text{Tm}^{-1}$, a drive-field strength of $20 \text{mT}^{-1}$, one drive-field patch is of size $40 \times 40 \times 20 \text{mm}^3$. Thus a full FOV of size $120 \times 120 \times 100 \text{mm}^3$ requires 45 focus-field stations. Assuming a drive-field repetition time of 1 ms and 1 ms for switching the focus-field stations one can thus image the full FOV within 90 ms, which should be sufficient for many cardiovascular applications.

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Figure 4. Reconstruction results of three different phantoms acquired at four focus-field stations. The first four columns show the separate reconstruction results when using only the system matrix and the measurement vector of a single focus-field station. The last column shows the joint reconstruction result.
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