Fast MPI Demonstrator with Enlarged Field of View

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Introduction

Magnetic particle imaging (MPI) is a new tomographic imaging modality that directly and quantitatively images iron oxide nanoparticle concentration without anatomical background signal [1]. It combines high sensitivity [2,3] with the ability of fast volumetric imaging. Current demonstrators either provide fast imaging or a large field of view. Here, a solution is proposed, that allows for both, fast imaging with large FOVs.

Methods

The current demonstrator (Fig. 1) has a bore size of about 120 mm. The selection field is generated by permanent magnets and has a strength of $dH_z/dz = 2.5\ T/\mu_0/m$ in vertical direction. The drive field is generated by three coils placed on the inner cylinder. The drive field frequency band is centered on 25 kHz and the maximum strength is $H_{Dx} = H_{Dy} = H_{Dz} = 20mT/\mu_0$. These performance measures closely resemble the setup described in [4], while the demonstrator improves bore size and exhibits a lower selection field strength. The smaller selection field strength increases the field of view accessible by the rapid drive field excitation to 16 mm x 32 mm x 32 mm. This, however, is not sufficient to cover a larger FOV, as allowed by the increased bore size.

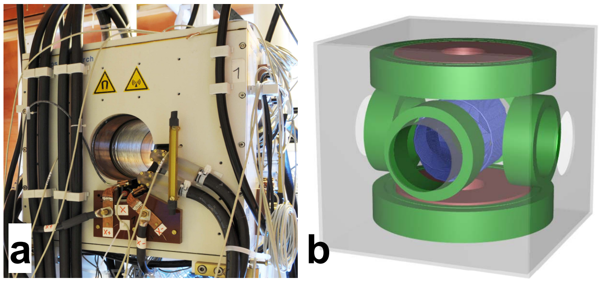

Figure 1: Second generation fast MPI demonstrator. (a) photograph (b) sketch of field-generating unit. Selection field magnets (red), focus field coils (green), and combined drive/receive coils (blue) are shown.

In order to overcome the limitations in FOV, an increased drive field strength could be used. For applications in the clinical domain, however, this might lead to an unacceptable heating of the patient. Therefore a third, additional set of coils, named focus field coils, is provided (cf. Fig. 1). They are able to provide $150\ mT/\mu_0$ in z (up-down) direction and $75\ mT/\mu_0$ in x and y direction, but operate at a much lower frequency than the drive field coils. The focus fields shift the area, which is scanned by the drive field, to cover a volume of $10 \times 10 \times 10\ cm^3$. To maintain fast acquisition, the focus field needs to shift the rapidly encoded area, named patch, by the linear dimensions of the patch within its encoding time. For an encoding time of more than 10 ms, the focus field strength needed to perform one shift is $40\ mT/\mu_0$, resulting in a slew rate of less than $4\ T/\mu_0/s$. This is well below the accepted field changes resulting from the gradient coil switching in a clinical MRI scanner.

In order to reduce the power consumption of the focus field, soft magnetic material was introduced to guide the focus field flux. This has the additional advantage of reducing the reactive power for a given slew rate. As the soft magnetic material is used mainly in the housing of the field generator “cube”, the stray field is shielded as well. Since soft magnetic material is known to generate harmonics that could interfere with the detection of the particles’ signals, it is shielded by aluminum plates. The thickness of the plates is chosen such as to prevent the drive field from penetrating the shielding, but to allow for transparency with respect to the low frequency focus field.

For this demonstrator, drive field signal quality and the background signal generated by spurious harmonics were analyzed quantitatively.

Results and Discussion

Harmonics generation and electromagnetic interference due to the drive field amplifier were found to be low enough for MPI operation. The peak power consumption for the focus field was 23 kW, leading to an average power of less than 6 kW for full FOV coverage. While this power level is feasible for this demonstrator, simply scaling the demonstrator to a size required for scanning a human would lead to several hundred kW of average power requirements. The main source of power dissipation is the z focus field coil. To overcome these limitations, it is proposed to use more soft magnetic material in future designs. In addition to the imaging capabilities, the possibility to move magnetic objects using the combination of selection and focus field could be demonstrated, which could be of interest e.g. for catheter steering in interventional radiology applications.

Conclusion

An MPI demonstrator capable of fast magnetic particle imaging in a large field of view was constructed, successfully employing the concept of slow focus fields for shifting the volume which is rapidly encoded by the drive fields. Using shielded soft magnetic material would allow for designs, which could be upscaled for clinical use.

References

[1] Gleich B, Weizenecker J. Tomographic imaging using the nonlinear response of magnetic particles. Nature 435:1214-1217 (2005).
[2] Weizenecker J, Borgert J, Gleich B. A simulation study on the resolution and sensitivity of MPI. Phys. Med. Biol. 53:N81-N84 (2008).
[3] Goodwill PW, Scott GC, Stang PP, Conolly SM. Narrowband MPI. IEEE Trans. Med. Imag. 28:1231-7 (2009).
[4] Weizenecker J, Gleich B, Rahmer J, Dahnke H, Borgert J. 3D real-time in vivo MPI. Phys. Med. Biol. 54:L1-L10 (2009).