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Multi-Channel Current Control System for Coupled Multi-Coil Arrays

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Abstract

For imaging and magnetic manipulation experiments in Magnetic Particle Imaging several field generating coils are required to produce sufficiently high and flexible magnetic fields. To minimize the power consumption, coils with iron cores are the best choice for low and medium frequency ranges. Such coils have comparatively high reactance and are often inductively coupled. The trivial approach to ensure target currents is to provide each coil with a current controlled source resulting in high system complexity and high costs. This paper presents a circuit design to distribute bipolar target currents from a single unipolar source with high accuracy, reducing unwanted coil coupling by a feedback controller. Thus, the number of current sources can be significantly reduced. With a regenerative concept, reactive power is stored and can be reused, allowing efficient and fast current switching.

1. Introduction

On the way to a human-sized Magnetic Particle Imaging (MPI) system several challenges need to be tackled [1–3]. One major issue is that limitations occur due to the up-scaling of all field generating coils. Sufficient magnetic-field strengths for selection and focus fields (SeFo) are hard to reach and can be restricted by infrastructure or cooling design. An optimized setup with soft iron or permanent magnets can lead to reduced power consumption of the system and help overcoming these challenges [4].

Efficient SeFo generators often are equipped with multiple iron core coils [5]. They have to be designed such that the ferromagnetic field enhancement properties of iron are effectively used. Using several coils allows for flexible magnetic field generation, which is essential for precise magnetic manipulation experiments together with simultaneous imaging [6]. Each coil built in the SeFo generator has to be supplied with its individual current. Depending on the operating scenario and the SeFo concept, currents should be allowed to flow bidirectionally.

Normally, this requires an expensive 4-quadrant amplifier for each coil. Additionally, current distortion due to coil couplings complicate stable current control.

In a real world setting, the total sum of currents per coil $I_{\text{system}} = \sum I_i$ within a SeFo sequence can be smaller than the sum of all maximum currents $I_{\text{max}} = \sum \max(I_i)$ needed and produced by individual current sources. In this paper, we present a design of a current supply for multiple coils that takes advantage of this circumstance. In essence, the concept is to use a single current source, which generates the current for all coils, instead of many individual sources. This requires a 1-quadrant amplifier that can provide the maximum of I_{system} instead of many 4-quadrant amplifiers. The current distribution to the individual coils can then be done via feedback controlled H-bridges, which receive a control signal from an IO unit. Depending on the number of coils M , acquisition costs can be reduced considerably and high currents can still be controlled with accurate precision. This approach is adapted for an 18-channel SeFo system driving coils with soft-iron cores.

II. Methods

The design and intended use of the SeFo generator results in various requirements for the current supply that need to be taken into account in the design phase.

II.I. Requirements

The maximum currents $\max(I_i)$ required for the 18 channel SeFo generator were calculated with an FEM software yielding up to 30A per coil. To allow imaging and force experiments with high temporal resolution, fast current switching should be possible. As a result the minimum current switching frequency is set to 10 Hz. The achievable frequency depends strongly on the connected coils, because for higher inductance more reactive power is required for current changes (here: $L_i \approx 1.3$ mH). Thus, the test case results in a sinusoidal current signal with 30A amplitude and a frequency of 10 Hz within the coils of the SeFo generator. Another requirement of the system is that the energy stored in the magnetic field must be handled. When the currents in the coils are reduced, the energy has to be dissipated or stored in an appropriate way.

II.II. Technical Realization

The current distribution circuit consists of three different parts. The first part is the supplying part, which is a current source together with a high power diode and a capacitor. The second part is the control unit, which includes the motor driver, a PID controller and an IO unit. The third part is the load consisting of the iron coil (L_i) and filtering components (C_r, C_p, L_r). A schematic layout of the circuit is shown in Figure 1.

For supplying all motor drivers with sufficient currents, two power supplies from Delta Electronics (SM 52-AR-60) are connected in parallel resulting in a maximum output current of 120A. Both sources operate in voltage controlled mode. To handle the energy stored in the magnetic fields, each motor driver directs the current from the coil back to the source when the current is lowered. Due to the high current diode, there is no current flowing back into the source and the energy from the coils is stored in a capacitor bank. When the current in a coil rises again, this energy can be reused. The capacity of the capacitor bank is chosen such that the total maximum energy can be stored without exceeding the maximum input voltage of the motor driver ($C = 92.8$ mF).

The control unit is realized by motor drivers from Sabertooth (Dual 60A 6V-30V Regenerative Motor driver) operating with a 24 kHz PWM-frequency. A single motor driver has two current channels, which can simultaneously supply a continuous current of up to 60A. A peak current of 120A is possible. To ensure current stability, an additional analog PID controller is integrated into the circuit, which measures the actual current and readjusts it if necessary. The current sensor in the PID controller

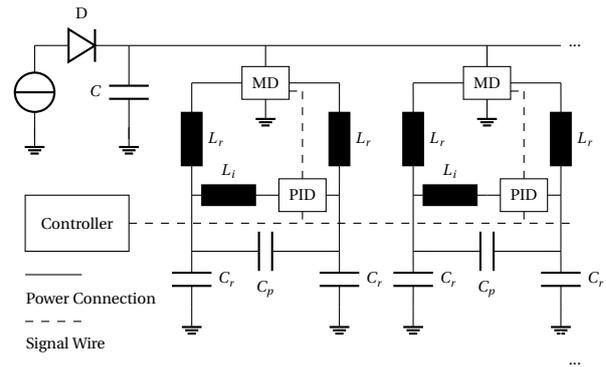


Figure 1: Current control system for M coupled coils (2 shown). All coils are supplied by individual motor drivers (MD), which are connected to one high current source. Each coil current in the field-generating coils (L_i) is set via a PID controller (PID). The set values for the individual current in each coil are generated by a cluster of RedPitayas represented by the controller box. With an additional capacitor bank (C) the reactive energy can be stored and reused. A diode (D) prevents the current from flowing back in the source. Current ripples are filtered by a series inductance L_r and capacitors C_r, C_p . Note that the number of motor driver circuits scales with the number of field generating coils being used.

limits the possible peak current to 110A and the continuous current to 50A. The setpoint of each controller is set by a cluster of RedPitaya IO boards running custom code from the open-source RedPitayaDAQServer¹ project. To provide the number of current channels a total of nine RedPitayas is required.

Additional current filtering by $L_r = 50$ μ H and $C_r = C_p = 10$ μ F is necessary, since the inductance of the iron coils strongly depends on the frequency. The 24 kHz ripple voltage from the H-bridge switching experiences a much lower inductance because of eddy currents damping the irons magnetization. As the iron cores are not laminated but made of solid soft-iron this effect can not be neglected. In addition, the filtering capabilities of an iron coil can be reduced at high currents, when its core is already saturated.

III. Results and Discussion

The current control system can supply multiple coils and is limited by the maximal output current of the supplying current source. In Table 1 some characterizing parameters are listed. The remaining current ripples are in the order of 300 mA and have a frequency of 24 kHz. For force experiments, high frequencies are averaged out, since the object can not follow such fast field changes. For imaging experiments, copper plates covering the coils would be necessary to suppress additional unsynchro-

¹<https://github.com/tnopp/RedPitayaDAQServer>

Maximum current (continuous)	50 A
Maximum current (peak)	110 A
Maximum frequency at 30 A	70 Hz
Ripple current at 30 A	± 300 mA
60 A switching edge time	<20 ms

Table 1: Current control system specifications.

nized particle excitation. These plates should be installed in any case, since the iron has to be shielded from the drive field in the imaging scenario.

The PID controller can precisely adjust the current output such that a 30 A sinusoidal current with a frequency of 10 Hz can flow in the coils according to the defined requirements. Even higher frequencies of up to 70 Hz are possible with 30 A amplitude. In the upper diagram in Figure 2, a comparison between the actual and the set values for a sinusoidal current are shown. Switching times for a 60 A current edge are about 20 ms.

Since multiple coils can share the same iron block, inductive coupling between the coils play a major role. To investigate this problem, one coil current is set to zero, while a nearby coil mounted to the same iron (see [7]) is driven with a 60 A current edge. Corresponding current measurements can be found in the lower diagram in Figure 2. The current curve for an uncontrolled and a controlled motor driver is recorded. In the uncontrolled case the current deviates by about 10 A from the set value. Furthermore, it takes more than 500 ms until the set value is reached again. With active controlling, the current changes by about 1 A and returns to its set value after less than 20 ms.

IV. Conclusion

In this work, we proposed a multi-channel current supply, which makes it possible to drive multiple inductively coupled coils with currents higher than 30 A and frequencies higher than 10 Hz. Current ripple and switching times are matching the requirements for the planned 18-channel SeFo generator. The proposed power supply design results in a significant cost reduction for setups with multiple coils that are individually driven with high currents.

Author's statement

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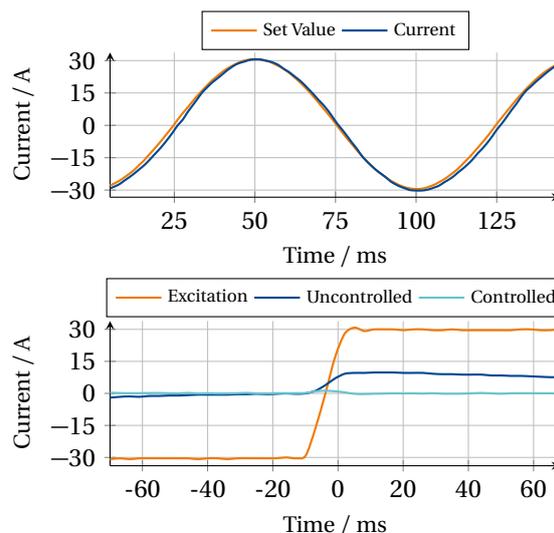


Figure 2: Upper diagram: Comparison between set and actual current for a 10 Hz sinusoidal waveform. Lower diagram: Influence of coil coupling. An exciting coil is experiencing a 60 A current edge. When the current in a nearby coil is not controlled long lasting current distortions up to 10 A occur due to coil coupling. With the PID controller the current is quickly adjusted to the desired value (here: 0 A).

References

- [1] J. Rahmer, C. Stehning, and B. Gleich. Remote magnetic actuation using a clinical scale system. *PLOS ONE*, 13(3):e0193546, 2018, Publisher: Public Library of Science. doi:[10.1371/journal.pone.0193546](https://doi.org/10.1371/journal.pone.0193546).
- [2] E. E. Mason, C. Z. Cooley, S. F. Cauley, G. A. Mark, S. M. Conolly, and L. L. Wald. Design analysis of an MPI human functional brain scanner. *International Journal on Magnetic Particle Imaging*, Vol.3:12 pages, 2017, Artwork Size: 12 pages Publisher: Infinite Science Publishing. doi:[10.18416/IJMPI.2017.1703008](https://doi.org/10.18416/IJMPI.2017.1703008).
- [3] M. Graeser, F. Thieben, P. Szwargulski, F. Werner, N. Gdaniec, M. Boberg, F. Griese, M. Möddel, P. Ludewig, D. van de Ven, O. M. Weber, O. Woywode, B. Gleich, and T. Knopp. Human-sized magnetic particle imaging for brain applications. *Nature Communications*, 10(1):1936, 2019, doi:[10.1038/s41467-019-09704-x](https://doi.org/10.1038/s41467-019-09704-x).
- [4] K. Sajjamar, J. Franke, H. Lehr, R. Pietig, and V. Niemann. Spatial selectivity enhancement in magnetic fluid hyperthermia by magnetic flux confinement. *International Journal on Magnetic Particle Imaging*, pp. Vol 7 No 1 (2021), 2021, Publisher: International Journal on Magnetic Particle Imaging. doi:[10.18416/IJMPI.2021.2103002](https://doi.org/10.18416/IJMPI.2021.2103002).
- [5] F. Foerger, M. Graeser, and T. Knopp. Iron core coil designs for MPI. *International Journal on Magnetic Particle Imaging*, 6(2):1–3, 2020, doi:[10.18416/IJMPI.2020.2009042](https://doi.org/10.18416/IJMPI.2020.2009042).
- [6] F. Griese, P. Ludewig, C. Gruettner, F. Thieben, K. Müller, and T. Knopp. Quasi-simultaneous magnetic particle imaging and navigation of nanomag/synomag-D particles in bifurcation flow experiments. *International Journal on Magnetic Particle Imaging*, pp. Vol 6 No 2 Suppl. 1 (2020), 2020, Publisher: International Journal on Magnetic Particle Imaging. doi:[10.18416/IJMPI.2020.2009025](https://doi.org/10.18416/IJMPI.2020.2009025).
- [7] F. Foerger, M. Boberg, M. Graeser, and T. Knopp. Low power iron selection and focus field generator. *International Journal on Magnetic Particle Imaging*, X(X), 2022, doi:[X](https://doi.org/10.18416/IJMPI.2022.2203076).