Investigating interactions between a TMS system and a novel MR device for concurrent TMS/fMRI experiments

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Target Audience

Researchers interested in concurrent TMS/fMRI studies and the potential interactions between MRI and TMS hardware.

Purpose

This work evaluates interactions of a novel dedicated RF coil array for concurrent TMS/fMRI studies [1] and a TMS coil (MRi-B91, MagVenture, Farum, Denmark) used for brain stimulation. The 3 T (123 MHz) MR head coil consists of a 7 channel receive only array, constructed on a spherical surface to allow free positioning on the head [2] (Fig.1). It can be placed between the TMS coil and the subject's head due to its ultra-slim design, ensuring efficient TMS stimulation and providing very high SNR. While the TMS coil itself is MR compatible and has been previously used in combination with a birdcage head coil, interactions between TMS and MR coil array need to be carefully evaluated due to their proximity and the high magnetic flux density routinely employed in TMS experiments (≈1 T at frequencies of about 1 – 10 kHz)

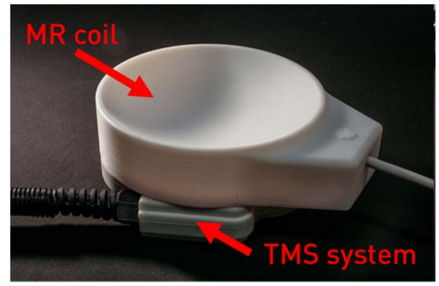

Fig. 1: Setup of TMS coil attached to the MR coil array.

Methods

Currents induced by TMS pulses on the array elements need to be negligible to avoid hardware (e.g. preamplifier) damage, TMS field disturbance and patient safety issues. The maximum induced voltage in a coil element by the TMS pulse was calculated using Faraday's law of induction on a wire loop coil of the same radius as the array elements, and consecutively measured using an oscilloscope with high input impedance (TDS3052, Tektronix, Berkshire, UK). For the analysis of induced currents, one coil element including decoupling and detuning circuits was simulated using the Advanced Design System software package (ADS, Agilent Technologies, Santa Clara, USA).
The previously measured induced voltage value was used in the simulation in order to calculate the current on the loop and the maximum voltage at the preamplifier input during TMS stimulation. The voltage at the preamplifier input was also measured.

The influence of the TMS coil on the RF characteristics of the coil elements was evaluated by measuring the shift in coil resonance frequency when attaching the TMS coil to the array.

The efficiency loss due to the increased distance between the patient and the TMS coil was quantified by mapping the TMS magnetic fields using MR phase images [3]. As standard TMS coil currents are too high for this approach, a dedicated low-current pulse generator supplied by the TMS Manufacturer was used to deliver 1 ms long block pulses to the TMS system in the range of 0 to 400 mA. Measurements were done on a Tim Trio MR scanner (Siemens, Erlangen, Germany) using a spherical phantom with the TMS attached, with and without the array between the phantom and the TMS device. Data was acquired with the body coil for both measurements, and also using the novel MR coil when it was placed on the phantom.

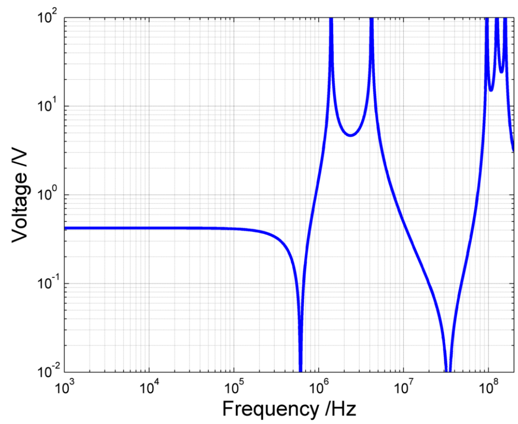

Fig. 2: Voltage spectrum at the preamp input calculated using ADS. Peaks between 1 and 10 MHz can be attributed to resonances of RF-chokes and DC-block capacitances. Voltages below 100 kHz are constant at 0.4 V.

Results

The voltage induced in a single loop by a 1 T amplitude pulse at 3 kHz was calculated as 53 V. Measurements using the simple wire loop and the TMS coil at 100% of its current rating resulted in a voltage of 33 V consequently used in circuit simulation, which showed a coil current of 9 μA and a voltage of 0.4 V at the preamplifier input (Fig. 2). Measurements showed a voltage of 0.7 V at the preamplifier input, dropping further to 50 mV after the preamplifier input capacitor. RF coil resonance frequencies were shifted up by maximal 1 MHz when attaching the TMS coil. TMS efficiency loss with the array between the TMS coil and the phantom was measured as ≈20% and is shown in Figure 3.

Discussion

Theoretical calculations and circuit simulation exhibited excellent agreement. The measured and calculated voltages and currents are not expected to cause any relevant TMS field distortions or hardware damage. They can be considered as worst case estimates, since the loop coil was placed directly under the TMS coil at the position with the largest flux density, whereas the actual setup includes a distance of at least 5 mm between the coil elements and the TMS coil. Furthermore, the central coil element closest to the TMS coil will not experience any significant induced voltage, since the surface normal flux density component is effectively zero close to the center of a figure-of-eight TMS coil. RF coil detuning due to the TMS is minimal and most likely due to a slight shielding effect of the extended TMS coil conductor copper structures. The TMS efficiency loss corresponds to the expected field decay at 7 mm depth, which is equal to the thickness of the coil array in the focus point.

Conclusion

Using a comprehensive set of simulations and measurements, we were able to validate that the TMS coil and the novel MR head coil for TMS/fMRI studies do not significantly interfere with each other in operation. The setup can thus be considered safe for patients as well as for the hardware. Using the TMS field distribution mapping method the loss of TMS efficiency could be accurately quantified. In future work, the coil array's thickness will be further reduced, resulting in even lower efficiency loss.

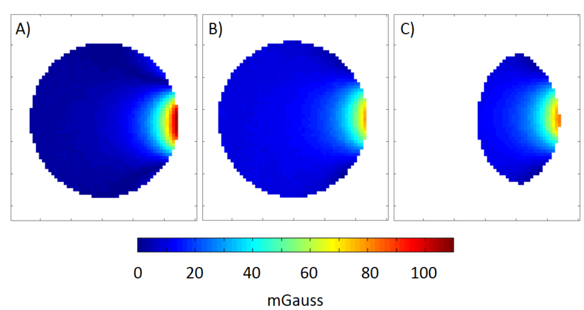

Fig. 3:TMS field map of a tangential component (x) with 150 mA TMS coil current. A) Without B)+C) with array coil between TMS and phantom. A)+B) Data was acquired with the body coil and C) with the coil array.

Acknowledgements

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References

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