Towards localization of transcranial magnetic stimulation inside the MRI scanner

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Background and Motivations: Transcranial magnetic stimulation (TMS) has been used as a non-invasive tool in various neuroscience applications including investigating brain functions, serving as a clinical diagnostic tool and as a therapy for several neuropsychiatric disorders [1]. TMS delivers time-pulse magnetic fields created by special coils to induce local currents in the tissue i.e., neuronal depolarization. Regardless of its expanding applications, TMS studies face the general problem of positioning the magnetic coil above relevant intended areas [2]. Currently, the standard localization is based on optically tracked frameless stereotaxic neuronavigation systems [1]. The precision of these methods are rather limited that the inconsistency of the coil placements may lead to the different results in repeated therapeutic studies [3]. In this work, we propose a new method of localizing the TMS stimulation by using MRI to map and visualize its magnetic field.

Materials and Methods: To localize the TMS field, we exploited the fact that the pulsed magnetic fields generated by the TMS coil interfere with the static field inside the MR scanner and lead to intensity variation of MR images. In principle, the TMS pulse triggered at any time between the excitation pulse and data acquisition would generate spatially varying field which can then be encoded using phase images. Unfortunately, high intensity of the TMS field (100% intensity corresponds to a maximum field of 2 Tesla, which is on the same order of magnitude with the static field) lead to a complete coherence loss of magnetizations and thus produce no MR signal. Here, the TMS pulse was delivered before each excitation pulse instead such that only the eddy current effect is observed while the field spatial information is assumed to remain unchanged. The delay between the two pulses was inserted with an aid of a Master-8 pulse generator (A.M.P.I, Jerusalem, Israel). Different delays and intensities were investigated to find an optimum set of parameter suited to localize the center of the field. The delay times used were 10, 11, 15, 20 and 25 ms for 10%; 10, 12, 15, 20, 25, 30 and 40 ms for 20%; and 10, 20, 30, 35, 40, 45 and 50 for 30% intensity.

All experiments were carried out in a spherical water phantom on a 3 Tesla Philips Intera scanner with a transmit/receive head coil. A standard gradient-echo EPI sequence with the following parameters was performed: TE = 30ms, TR = 2s, flip angle = 90º, FOV= 150x150 mm^2, matrix = 128x128. Data from twenty 1mm-axial-slice across the center of the phantom were obtained and both magnitude and phase images were reconstructed. An MR compatible figure-of-eight TMS coil (Magstim®, Wales, UK) was laid flat on the bed underneath the phantom with the stimulator side facing up mimicking a position corresponding to stimulation of the visual cortex. Further, to investigate the extent of TMS pulse as a function of its intensity, differences in signals were calculated from the corresponding magnitude images. The ROI used here (see Fig. 1a) were chosen to include the affected area without the aliasing.

Results and Conclusion: Figure 1 show phantom images when TMS was applied with a 10 ms delay at different intensities. Some geometric distortions were also observed in the images which are likely caused by the presence of eddy current during data acquisition as the distortions mitigate with longer delay times (data not shown). Phase images (Fig. 1 f-h) visibly illustrate the magnetic field to be originated from the left and right side of the coil (red arrows). Differences in image intensities as a function of the delay time shown in Fig. 2 displays decreasing effect with increasing time as expected. The non-zero asymptote may arise from geometric distortion mentioned above since the eddy current has been reported to last as long as 100 ms after the TMS pulse [4].

Our work demonstrated that the origins of the TMS fields could be localized with a simple MR imaging technique. Precisions of the method although not validated here will be on the same order of magnitude as that of the resolutions of the acquired images. Further modifications to improve our technique include eliminating eddy current effect during data acquisition by employing eddy current compensation gradients. Mapping of the TMS fields with such technique can potentially lead to quantification of neurostimulations that the TMS produce.

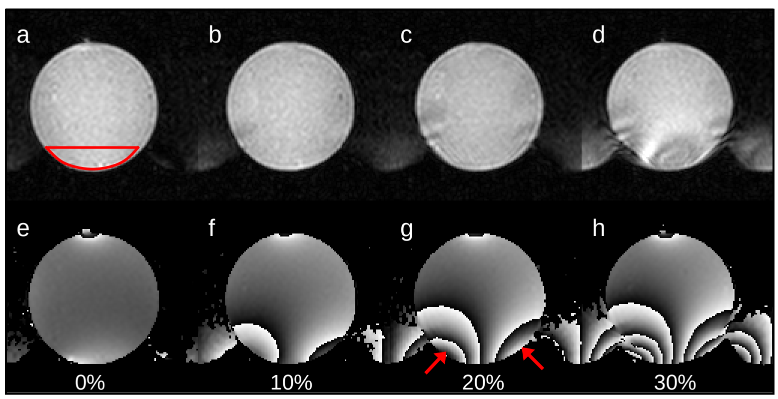

Fig. 1 Magnitude (a-d) and phase (e-h) images with TMS applied at a 10ms delay for 0, 10, 20 and 30% intensities. Arrows indicate center of magnetic field generated by the TMS coil. Unwrapped phase images simply display greater field distortion (more concentric rings) with higher TMS intensities.

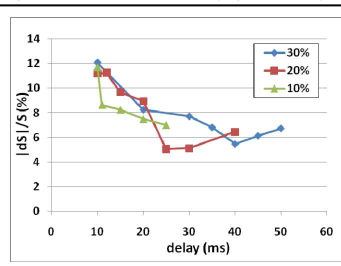

Fig. 2 Differences in MR signals show that higher intensity TMS pulse requires longer delay time to reach an asymptotic value.

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