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Application of HARDI to assess optimal coil orientation in neuronavigated TMS of the motor cortex

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Introduction

Transcranial magnetic stimulation (TMS) is a modality for noninvasive brain stimulation with rapidly growing diagnostic[1] and therapeutic[2] applications. While used in the treatment of conditions such as depression or neuropathic pain, results of neuromodulation protocols are known to be heterogeneous between individuals and centers[3]. Additionally, the neurophysiological processes underlying TMS effects are still insufficiently understood[3]. One factor that might contribute to optimized results of TMS is the orientation of the stimulating coil in relation to cortical and subcortical anatomy. We present preliminary results from healthy participants in whom we investigated the role of intragyral fiber orientations in terms of TMS effects on the motor cortex.

Methods

20 healthy participants (average age: 27 years, 10 females) underwent high angular resolution diffusion imaging (HARDI; 60 gradient directions modeled on a sphere [b-value=1500 s/mm^2] and 6 interleaved b0 volumes) and T1-weighted (T1w) imaging at 3 Tesla. The T1w images were used for neuronavigated TMS (nTMS). Up to three nTMS sessions separated by at least 14 days were conducted per subject to assess the robustness of the optimal coil orientation. In each nTMS session, we elicited 140 motor-evoked potentials (MEPs) from the abductor pollicis brevis muscle hotspot using 7 different coil orientations (30° - 150° relative to the longitudinal axis of the precentral gyrus, in 20° steps) on the dominant hemisphere. The MEPs were analyzed regarding the influence of coil orientation on MEP amplitudes. Additionally, HARDI data were corrected for signal drift, motion, echo planar imaging/EPI), and eddy current (EC) distortions and coregistered to the T1w imaging used in the navigation of TMS via ExploreDTI[4]. Herein, constrained spherical deconvolution (CSD) truncated at maximum harmonic order L-max=8 was used to model fiber orientation distributions (FODs). These were visualized in relation to the observed optimal stimulation direction (Figure 1).

Results

In total, 6720 individual MEPs were analyzed. The MEP amplitude correlated significantly with coil orientation (rho = -0.39, p << 0.0001; Figure 2). A random effects linear regression model predicted a deviation from the optimal orientation (defined by highest mean MEP amplitude) by 20 degrees to lead to MEP decreases of -168 µV (β=-168, t=-28.97). On a qualitative level, when comparing the optimal stimulation direction to HARDI-derived fiber orientations, we noticed a notable parallel alignment of FODs to the e-field direction yielding maximal MEPs (Figure 1).

Conclusion

Coil orientation during TMS in relation to gyral anatomy significantly modulates motor responses. HARDI-derived fiber orientation imaging could aid in predicting optimal coil orientation in non-motor areas, e.g. for therapeutic TMS applications.

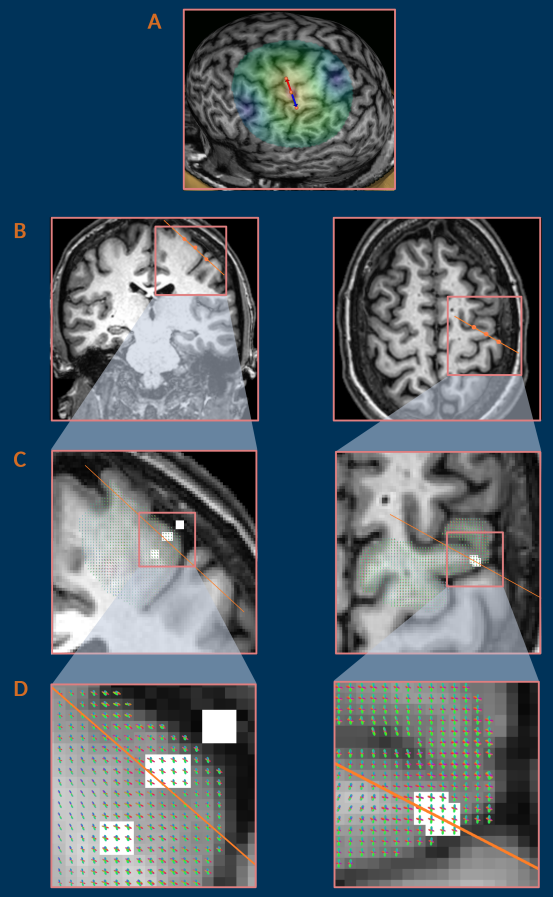

Figure 1: A) Optimal e-field direction was identified via highest mean motor evoked potential (MEP) amplitude (arrow). B) Projection (orange) of optimal e-field direction in axial and coronal planes. C) HARDI was coregistered to the T1-weighted (T1w) imaging data (white dots indicate maximum e-field at different depths within the brain). Fiber orientation distributions (FODs) were modeled for each voxel, representing main diffusion directions along axonal orientations. D) The optimal e-field appears to run parallel to one of the dominant FODs. This was interpreted as reflecting maximal excitability when the e-field is oriented parallel to axonal structures.

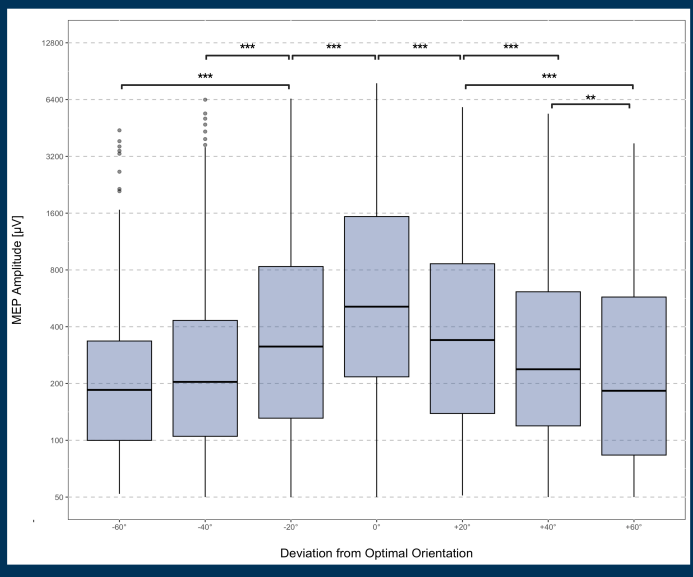

Figure 2: MEP amplitude changes depending on e-field orientation in relation to the optimal direction (0°, middle plot), which was defined as eliciting the maximum mean ME amplitude out of the tested coil directions. The ME amplitude was lowered significantly by deviations as little as 20° from the optimum.

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