On the electrodynamic constraints and antenna array design for human in vivo MR up to 70 Tesla and EPR up to 3GHz

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Target audience: Basic researchers and clinical scientists interested in ultrahigh field magnetic resonance and in-vivo electron paramagnetic resonance.

Purpose: Neuroimmunological, neurovascular and neurodegenerative diseases benefit from enhanced diagnostic information exploiting the signal-to-noise (SNR) gain inherent to ultrahigh field (UHF, $B_0 \geq 7T$) MR [1]. While today's lion's share of brain UHF-MR covers anatomical imaging, more physiological characterization of brain tissue including O_2 metabolism and tissue pO_2 are other directions that can benefit from higher magnetic field strengths. O_2 characterization in brain tumors of small animals can be achieved with electron paramagnetic resonance (EPR) imaging, which utilizes intrinsic high sensitivities of an electron gyromagnetic ratio that is 658 times superior to that of protons [2]. While conceptually appealing in their application range, UHF MR and EPR share the suffering from electrodynamic constraints dictated by an increase in spin excitation frequency rendering human in vivo applications challenging if not elusive. Power losses due to a frequency dependent increase in conductivity reduces B_1^+ efficiency and increases the specific absorption rate (SAR). In this light appropriate RF antenna design may offset some of the SAR predictions governed by SAR ~ $B_0^2 B_1^2$. Recognizing these opportunities, this work utilizes electromagnetic field (EMF) simulations in phantoms and human voxel models to analyze transmission field (B_1^+) and SAR distributions of multi-channel antenna arrays in the frequency range of 300-3GHz corresponding to an effective field strength of 7-70T for proton MRI or ~10-100mT for EPR. The results and conclusions drawn are relevant for in vivo UHF MR and in vivo EPR.

Methods: EMF were performed with CST MWS at discrete frequencies ranging from 300MHz to 3GHz. A virtual phantom study was implemented on a cylindrical phantom (r=90mm) with frequency adjusted brain tissue properties (60% white matter + 40% grey matter). B_1^+ and SAR were calculated for a 1- and 8 channel (ch) dipole antenna array positioned symmetrically around the phantom. In a second step the number of transmit channels was increased at the given frequency to a maximum number of channels (maxCh) under the condition that decoupling doesn't increase above -13dB. Absolute $B_1^+/\sqrt{kW}$, SAR (point,10g,1g) and $B_1^+/\sqrt{kW}/\sqrt{maxSAR}$ were compared at different frequencies. EMF simulations were performed with a 20ch dipole antenna array at 1GHz and a 25ch dipole array at 2GHz and 3GHz positioned around the head of the human voxel model "Ella" from the virtual family with frequency adjusted tissue parameters [3]. Validation experiments were performed in RF heating experiments in custom built phantoms at 300 and 500MHz.

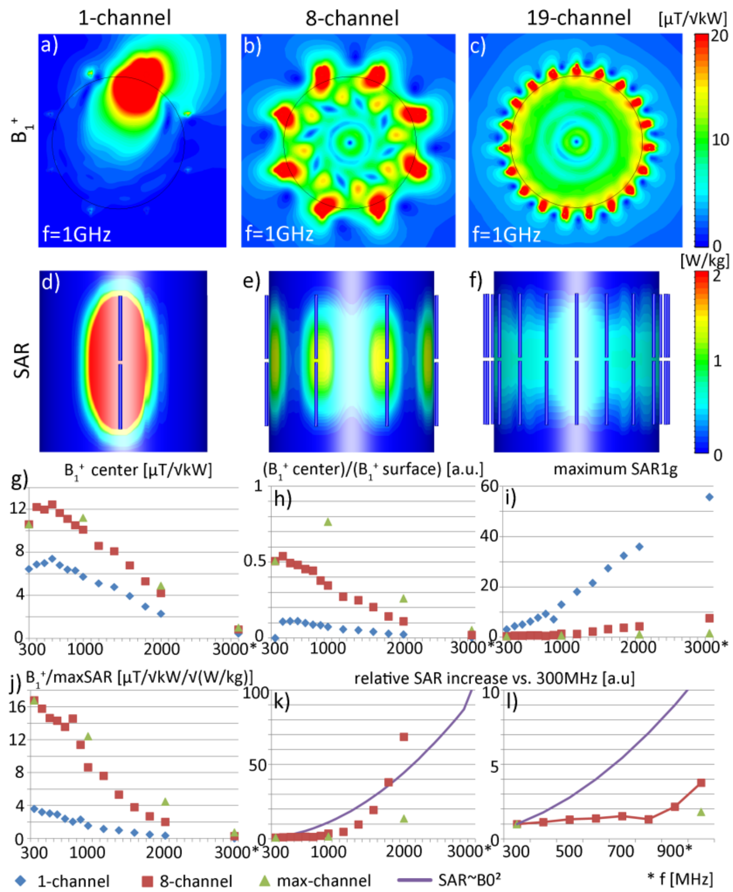

Figure 1 a-g) B_1^+ and SAR distribution at 1GHz for a 1-channel, 8-channel and 19-channel antenna array. **g-l)** B_1^+, SAR and relative SAR increase to reach the same flip angle for frequencies ranging from 300MHz to 3GHz.

Results: A higher number of elements improves effective B_1^+ in the center of the phantom, while reducing SAR (Fig.1a-c & Fig. 1d-f). B_1^+ in the center of the phantom decreases with an increase in frequency, while maximum SAR1g increases (Fig.1g-i). A shorter wavelength at higher frequencies results in lower decoupling between the antennas. While at 300MHz only 8 elements can be positioned around the phantom to condition decoupling values <-13dB, at 1GHz/3GHz the number of elements can be increased to 19/22. Such arrangement improves at 1GHz (Fig.1c,1f) maximum B_1^+ by 11% while decreasing SAR1g by 41%. The ratio of center-to-surface B_1^+ was improved from 0.35 to 0.77 by using 19 vs. 8 antennas indicating an increase in B_1^+ homogeneity. Taking frequency dependent B_1^+ decay and SAR increase into account the relative SAR increase vs. 300MHz was found to be below SAR~B0² for 8-channels and f<1.8GHz (Fig. 1k). If the advantages of higher RF frequencies are exploited, the relative SAR increase can be reduced significantly. At 2GHz this would result in a factor 10 using 22 elements instead of a factor 70 SAR increase when transmitting with 8 antennas (Fig. 1k). At 1GHz, which corresponds to a proton excitation field of ~23T, SAR increases by only 1.8 as compared to 7T (Fig.1k-l). The human voxel model simulations confirm these findings. In comparison to bow tie antennas tuned to 300MHz and exhibiting similar length [4], the relative SAR increase for a 20ch dipole array at 1GHz was found to be 1.8 (Fig. 2).

Discussion: A shorter RF wavelength inherent to higher RF frequencies benefits an application of phased array antennas, in particular electric dipoles with their small dimension in the axial plane. Improved decoupling allows for a high transmit channel density around the human head leading to more equally spread RF power over the surface while constructive summation of each propagated wave increases B_1^+ in the center of the human head.

Conclusion: Notwithstanding the challenges dictated by electromagnetic constraints and practical obstacles for human in vivo MRI and EPR utilizing frequencies larger than 300MHz, the benefits of multi-channel antennas at higher frequencies render MRI at 23T and EPR at 1GHz and above feasible from an electrodynamic standpoint. This finding is in positive alignment with recent explorations into 20 Tesla human magnet design [5].

References: [1] Kuchling J, et al., Nervenarzt, 2014 [2] Eaton S and Eaton G, JMR, 2012 [3] Hasgall PA, et al., IT'IS database, 2014 [4] Winter L, et al, PlosONE, 2013 [5] ISBN: 978-0-309-28634-3, The National Academies Press, 2013

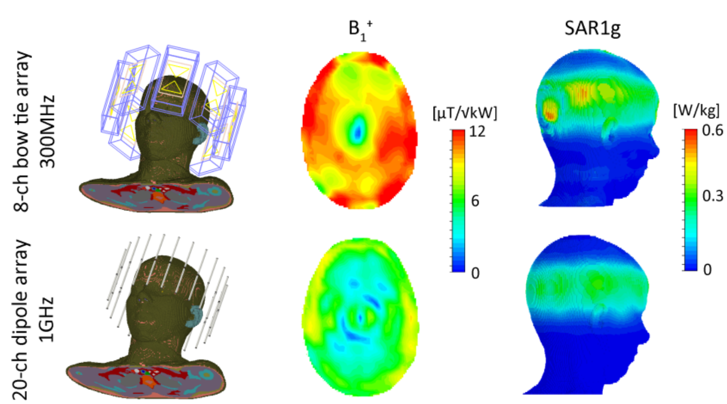

Figure 2 Comparison of B_1^+ and SAR1g simulations for (top row) an implemented 8-channel bow tie array at 300MHz [4] and (bottow row) a 20-channel dipole array at 1GHz.