The ultimate SNR and SAR in realistic body models

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Target audience: RF engineers and MR physicists.

Purpose: The ultimate SNR (uSNR) (1-3) and SAR (uSAR) (4,5) have been computed for simple sample geometries like spheres and cylinders for which analytical basis of the solutions to Maxwell equations exist. These calculations provide valuable insight about how well a given design is performing and the potential for gain via further effort. However, RF coils are designed for the human body, not cylinders and spheres, and the conclusions of this work on ultimate SNR and SAR work are weakened by this restriction to simple geometries. Characterizing a modern head-neck array using a spherical phantom provides a poor metric since many elements are improperly loaded and some regions (such as the neck) cannot be evaluated. In this work, we compute for the first time the uSNR (accelerated and un-accelerated) and the uSAR in a realistic non-uniform body model ("Duke" IT'IS Foundation, Zurich) at 7 T. This is made possible by a recent breakthrough in ultra-fast electromagnetic (EM) simulation in the form of a volume integral solver that allows computing EM fields in complex non-uniform body models with minutes [8].

Methods: Computation of an EM basis for the "Duke" body model: Computation of the uSNR and uSAR require a basis of the solutions to Maxwell equations in the sample studied. We compute such a basis in the Virtual Family "Duke" upper torso body model (7) at 7 T in three steps follows. STEP #1: We define a surface on which we allow arbitrary current patterns to flow. This surface encloses the body model and is located at a distance D from it (D=3 cm in this work). We place a large number (~10^6) of electrical dipole sources on this surface pointing in the x, y and z directions. STEP #2: We randomly excite common modes of these dipoles (random phases and amplitudes). We compute the E and B fields of these excitations in the body model using an ultra-fast full-wave electromagnetic solver (~45 sec. per solve) (8). STEP #3: Using a randomized SVD, we compute the relative singular value drop associated with the basis vector produced in the previous step. We stop adding vectors to the ultimate basis when this value falls below 1%. Computation of the ultimate bounds: We calculated the un-accelerated uSNR in the Duke body model as described in (3). Similarly, we computed ultimate G-factors for Cartesian SENSE imaging by treating the vectors of the ultimate basis as the "coils" (B1-maps) (9) (the correlation matrix was computed using the E fields of the basis vectors, i.e. resistive coupling was modeled but the mutual inductance between the coils was assumed to be zero). We also designed parallel excitations RF pulses targeting a uniform 30° excitation pattern using the global SAR matrix and the B1+ maps associated with the ultimate basis. Least-squares RF shimmings. pulses were designed using a fast spokes design algorithm with explicit global SAR limit (10) allowing analysis of the global SAR vs. excitation fidelity tradeoff. For comparison, we also computed the SNR of three 32, 64 and 128 channels receive coils and the global SAR vs. excitation error performance of two 8 and 16 channels transmit coils (we used our ultra-fast EM solver for this purpose and no inductive coupling was modeled between the coils).

Results/Discussion: With the computers available, we were able to compute 6,000 basis vectors in the Duke body model (more RAM is needed to go further). Fig. 1a shows that this is enough to accurately estimate the uSNR at most positions on the brain, although adding more basis vectors would likely improve the accuracy of the estimation at the edge of the brain. Fig. 1b and 1c show that the SNR of the 128 channels coil is within 20% of the optimum at all positions in the brain except at the very edge of the grey matter. For accelerated imaging (Fig. 2), we observe, like other authors previously in sphere models, that all coils are almost optimal at R=1x4 and R=2x4. The performance of the 32 and 64 channels coils degrades rapidly for R=2x4, but interestingly the 128 channels is almost optimal. This indicates that the use of additional channels beyond 128 at 7 T may not be advantageous for brain imaging (more coverage may be wanted for imaging the neck however). Fig. 2 also shows that the ultimate G-factor is not equal to 1 at high accelerations, which indicates a fundamental limit of parallel imaging at 7 T (this is in agreement with the results of Wiesinger et al (6)). Fig. 3 indicates that 8 and 16 channels parallel transmit coils (most commonly used channel counts) are far from the ideal performance, where a simple least-squares RF shimming pulse design strategy can achieve almost perfect flip-angle excitations at extremely low global SAR .

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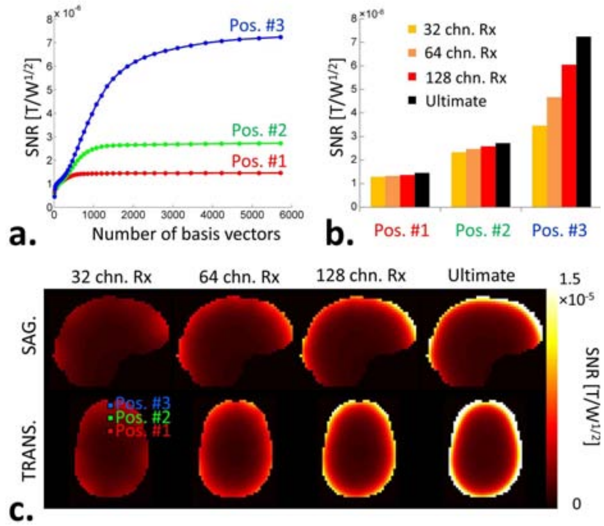

Fig 1. a: Convergence of the ultimate un-accelerated SNR at 3 positions in the brain of the Duke body model as a function of the number of vectors in the ultimate basis at 7 T. **b**: SNR of three 32, 64 and 128 channels receive coils compared to the ultimate at 3 positions in the brain. **c**: SNR maps.

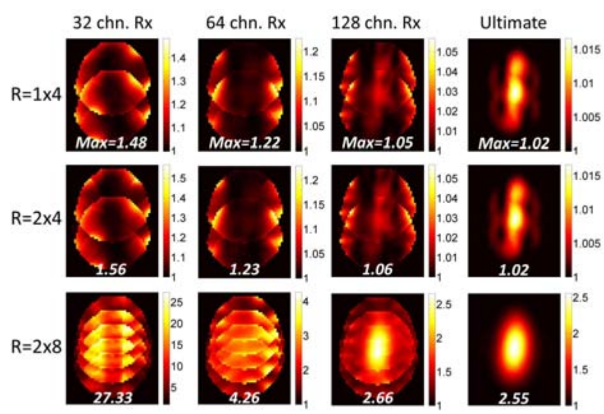

Fig. 2. Cartesian SENSE G-factors for three 32, 64 and 128 channels receive coils and the ultimate basis in the brain of the Duke body model at 7 T.

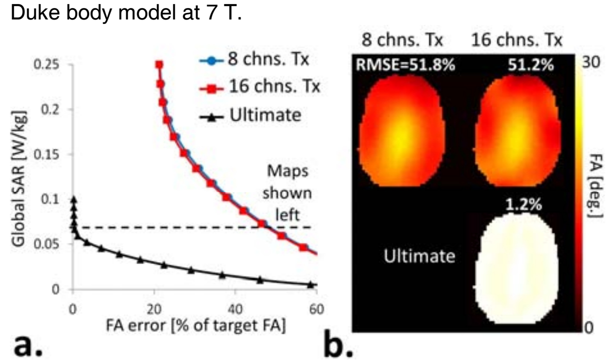

Fig. 3. a: Tradeoff between global SAR and excitation fidelity for two 8 and 16 channels transmit coils and the ultimate basis in Duke at 7 T. **b**: Flip-angle maps at constant global SAR (0.06 W/kg) for the 8 and 16 channels coils and the ultimate basis.