SAR characterisation for parallel transmission MRI – comparison between modelling different decoupling regimes

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Target Audience: This work will benefit those interested in modelling parallel transmission MRI systems for SAR characterisation.

Purpose: Accurate characterisation of SAR is crucial in parallel transmission MRI due to the variation in location of local SAR hotspots. Effective modelling of a system enables control over SAR and can be used to adjust the system drives with the aim of reducing maximum local SAR[1]. However modelling parallel transmit coils is complex due to potentially strong coupling between transmission elements. Small differences between simulation and experiment lead to variations in simulated lumped element values required to match the physical system behaviour, so tuning the simulated system is troublesome; particularly since running EM field solver software is time consuming. A commonly used solution is to model the individual transmit elements independently[2] without including any decoupling networks, which themselves would also require optimising. This leads to an "ideally decoupled" system model, which is relatively simple to implement. Alternatively, circuit co-simulation[3] can be used to optimise the lumped elements in the much less computationally demanding circuit domain. This enabled values of each capacitive element to be iterated in an optimisation without running a full-wave EM simulation bringing the computation time per iteration down to the order of milliseconds rather than days, allowing the entire system to be modelled. However these simulations are considerably more complex to implement. We shall refer to this method as the "fully modelled" simulation. In this study we compare the two methods.

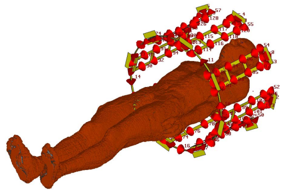

Fig. 1: Model configuration

Methods: The system modelled was a 3T Philips Achieva MRI scanner fitted with an 8-channel body transmit coil[4]. Simulations were performed using the time domain Finite Integration Technique of CST Microwave Studio (CST AG, Darmstadt, Germany). All lumped elements were modelled as 50 Ω S-parameter ports (Fig. 1) and the NORMAN male voxel model[5] was used with a heart centred configuration. To optimise the capacitor values in the decoupling network[4], S-parameters from all ports were exported to Matlab and optimisations were set up based on minimising the S-parameter matrix at 128 MHz in order to tune, decouple and match the coils[6] (Fig. 2). The minimisation, $\arg\min\{\ \|S_{i=j}\| + \lambda(\max\{\ |S_{i\neq j}|\ \})\ \}$ included a parameter λ to tune the optimisation to favour either the matching or nearest-neighbour coupling of the network. The S-parameter ports had the resulting optimal capacitor values applied to them in the circuit simulation and the corresponding E and H-fields were created. The ideally decoupled system was modelled by setting all capacitors in other coils to values of $\sim10^{-50}$ F (effectively removing them) and then running the circuit simulation for each transmit element in turn. The fields and voxel model were exported to Matlab and interpolated onto an isotropic, hexahedral 5mm grid. Q-matrices were calculated for both sets of electric fields, which were then condensed down into a set of Virtual Observation Points[7] (VOPs) with a 1% overestimate bound for SAR comparison. The fully modelled fields contain residual coupling between channels as would be the case in the physical system. In order to compare the fields with the ideally decoupled case, a linear combination transformation was calculated to actively decouple[4] the fully modelled fields. This active decoupling matrix was found by fitting the fully modelled B_1^+ fields to their ideally decoupled counterparts. The resulting matrix was applied to the corresponding E-fields.

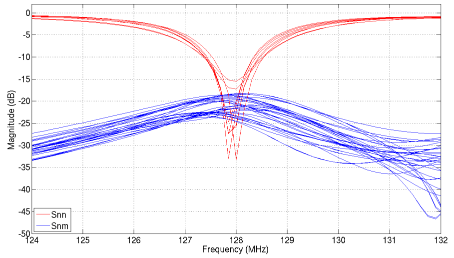

Fig. 2: S-parameters plotted showing matching and decoupling equivalent to the physical coil (S_{nn} <-15 dB and S_{nm} <-18 dB at 128 MHz)

Results: The electric fields from the two models show differences in the regions close to other transmit elements (Fig. 3a,b). This is to be expected as the ideally decoupled case does not model the capacitive components in other transmit elements. Within the body however, the similarity between the two models is greater as can be seen by the difference maps in Fig. 3c. The maximum variations are primarily seen in the elements which are strongly loaded by the voxel model's arms – the ideally decoupled simulation deposits more power into the voxel model. These variations are seen to have a smaller impact on maximum local SAR estimates (Fig. 4). Comparing maximum local SAR estimates for a set of 100 random, complex drive settings using VOPs from both methods against the full Q-matrix set from the fully modelled simulation shows that the ideally decoupled simulation leads to overestimates of up to 6%.

Discussion: It is clear that the ideally decoupled simulation leads to overestimates of SAR. This is likely due to greater power dissipation in the subject as there are no other active coil elements for the power to dissipate into elsewhere. However the SAR increase is relatively small and this may further be offset by the fact that the ideally decoupled simulation is much simpler to optimise as there are far fewer variables involved for each transmit element, and the optimisation is more stable due to the lack of coupling. Furthermore, the ideal EM simulation would be faster to run as the decoupling network could be ignored leading to fewer ports to be excited (in our model this would produce only a modest reduction in simulation time of 12.5%).

Conclusion: We have shown that running ideally decoupled simulations in place of fully modelling decoupling networks can produce comparable results. Ideally decoupled models are simpler to set up and run, however in our tests they resulted in systematically larger estimates of SAR. This would lead to conservative limits on scanning which may not be ideal if the user aims to use parallel transmission MRI at the SAR limits. In contrast, full modelling, as described here, provides effective means of characterising the whole system with computational times that are <15% longer. Such approaches may also potentially provide greater insight into the systems being modelled.

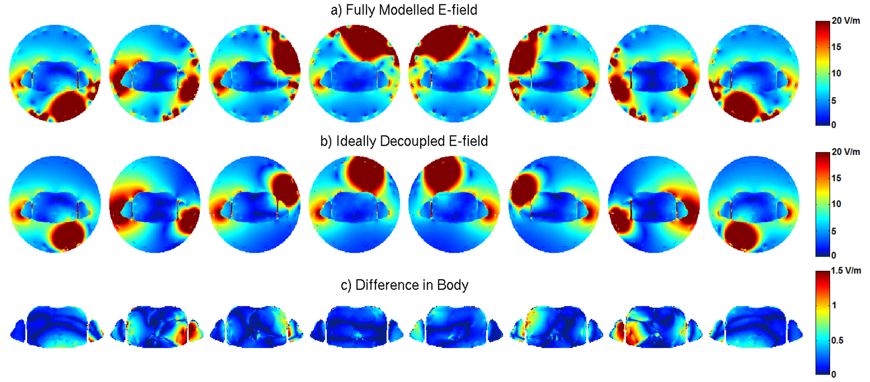

Fig. 3: a) and b) show the absolute E-field magnitude for the fully modelled (with active decoupling applied) and ideally decoupled simulations respectively and c) shows the absolute difference (ideal - full) between these in the body – N.B. different colour scale

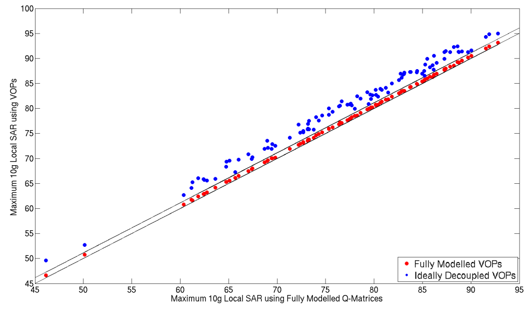

Fig. 4: Plot of VOPs from both simulations compared with fully decoupled Q-matrices – overestimates of up to 5% can be seen in the ideally decoupled VOPs

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