Considerations for using linear combinations of array elements in B1 mapping

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Introduction: Accurate B_1 mapping of transmit coil arrays is vital for parallel transmission MRI. Current mapping methods accommodate a limited range of B_1 amplitude, but can be adapted for use with coil arrays by mapping B_1 fields from linear combinations of coils and solving for individual elements [1,2]. Existing formulations can be represented by an encoding matrix A, with $A_{i,j} = 1$ for i≠j, and $A_{i,j} = \varepsilon$ for i=j. Setting ε=0 gives [1]; ε=-1 gives [2] and ε very large gives the single driven coil case. The possibilities for choosing A are vast, and the optimal choice is likely related to the mapping technique as well as the coil array/loading in any particular case. In general we require that the resulting linear combination B_1 maps have a low dynamic range of flip angles but also that the condition number of the transformation is low so that noise amplification upon inversion is minimised. We have studied the performance of the family of encoding matrices generated by varying the parameter ε using numerical simulations with real coil data from a parallel transmit system, when used with the AFI B_1 mapping technique [3].

Method: All scanning has been performed on a 3T Achieva MRI system (Philips Healthcare, the Netherlands) equipped with an 8 channel body coil capable of parallel transmission [4]. Single slice B_1 maps were acquired using AFI from four different targets; the pelvis and brain of healthy volunteers, a 400 mm doped water phantom and a 400 mm mineral oil phantom. Signal averaging and the technique from [1] were used in order to obtain good-quality maps resulting in scan times of approx 11min in each case. Note that the array is configured such that driving all coils equally results in nominally quadrature excitation.

For each dataset, the parameter ε was varied in order to generate different encoding matrices (A). Linear combination B_1 maps (B_1^{LC}) were calculated by transforming the coil B_1 maps with A, and renormalised to achieve a constant maximum B_1 value. The renormalisation means that ε controls only the relative drive applied to each channel. Using signal relations taken from [3], simulated gradient echo images from the AFI technique were generated using a peak flip angle of 60°, and TR_1=25ms, TR_2=125ms. A small amount of noise was added (σ≈5% of maximum signal) and then B_1 maps were derived from the ratio of these images. The discrepancy between the derived maps and the original linear combinations formed from the gold standard maps can be thought of as the "mapping error". These maps were then subjected to the inverse transformation of A to yield estimates of the individual coil B_1 maps. The difference between these maps and the original gold standard data is the total error. Both of these error metrics were calculated using normalised root mean square (RMS) difference. For all B_1^{LC} a measure of the dynamic range was calculated as the RMS deviation from uniformity.

Results: Figure 1 shows the condition number of the transformation and dynamic range of the resulting linear combinations for ε in the range -5 to 9 as well as the mapping error and total error for each of the four scenarios. Strikingly the mapping error closely follows the form of the dynamic range measurement, whereas the total error is clearly also influenced by the condition number of the transformation; both as would be expected. For this parameterisation there are two cases where cond(A)=1. The first is ε→±∞ (single coil driven) and the second is ε= -(Nc/2 -1), where Nc is the number of channels. The second choice results in the scalar of product of any two rows of A being zero; for Nc=8 this gives ε=-3. Figure 2 shows example B_1 maps and errors for selected values of ε. From fig 1 it is clear that the optimal transformation depends strongly on the dynamic range, and is therefore different in each case. There are however some common features, for example ε=1.1 results in very low mapping error (and dynamic range) in all cases but very high total error because the condition number is 81. For large fields of view (pelvis and both phantoms), ε<0 results in focal regions where B_1^{LC} ≈0 due to cancellation between adjacent elements driven with opposing phase. For the oil phantom, this problem is limited in extent since the fields from each coil decay quickly away, however for the water phantom and human pelvis the effects are more severe. In the case of the water phantom it means that the overall optimal ε≈3; a trade off between mapping error and condition number. For the brain however the results are quite different. The fields from each coil do not vary much across this small field of view in the centre of the array. As a result ε<0 does not result in strong signal cancellation for smaller values, and in fact the optimal choice in this case is ε≈-2.5.

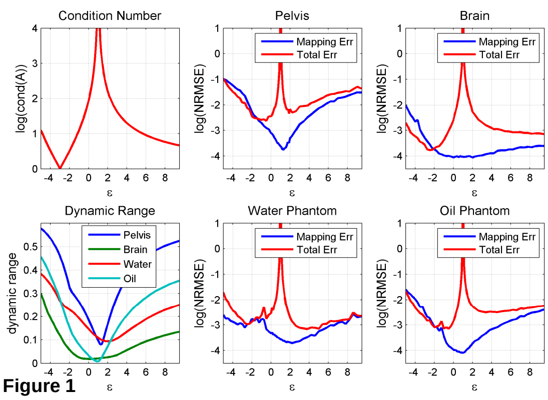

Figure 1

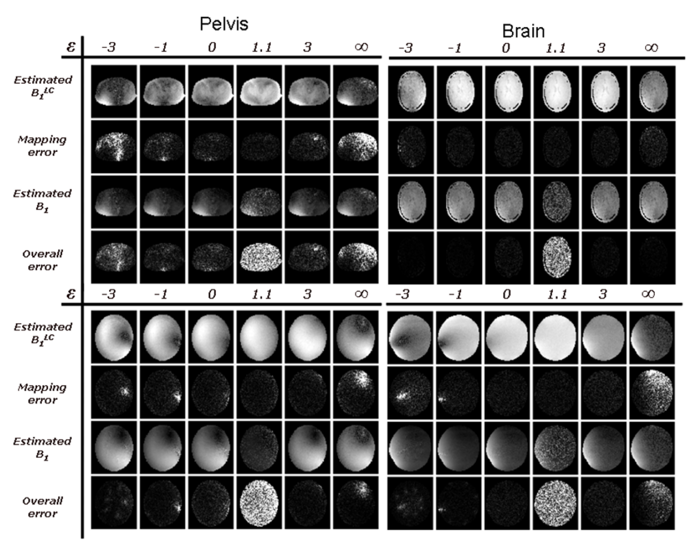

Figure 2

Discussion & Conclusions: The results indicate that the optimal transformation depends on the size and loading properties of the object. In general choosing ε<0 leads to lower condition number and therefore less noise amplification, and ε=-3 is optimal if there are no regions in which the fields cancel. For larger fields of view this is not possible to achieve, and so a trade off between condition number and dynamic range is necessary. The choices ε=0,-1 [1,2] result in different balances of these in line with the above. The parameterization used in this study is very simple, with only one free parameter. In fact there are many more degrees of freedom available, and choice of more general linear combinations may result in better results, however we have not yet found a systematic way of assessing the general case. Finally, the normalisation used affects the results obtained. We normalised to maximum B_1, which avoids producing very strong local fields, and also ensures an effective use of the dynamic range available for mapping without exceeding the maximum flip angle for which the technique works.

References: [1] Nehrke et al, ISMRM'08 #353 [2] Brunner et al, ISMRM'08 #354 [3] Yarnykh, MRM 2007;57 [4] Vernickel et al, MRM 2007;58