Very Fast Multi Channel B1 Calibration at High Field in the Small Flip Angle Regime

P-F. Van de Moortele[1], and K. Ugurbil[1]

[1]Center for Magnetic Resonance Research, University of Minnesota, Minneapolis, MN, United States

Introduction: B1 inhomogeneity is a challenging obstacle in human MR experiments at high field. Static B1 shim or Parallel Transmission approaches can address this issue, but such techniques require mapping the transmit B1 profile (B1+) of each transmit channel. This is challenging with a large number of coils and in a large volume. Furthermore, most B1 mapping techniques require either high peak power to obtain a sinusoidal variation of Mz with RF amplitude or long repetition times to avoid T1 bias. Here we introduce a very fast method to estimate B1+ in transceiver arrays based solely on fast, small flip angle Gradient Echo (GRE) images. We compare this approach with a large flip angle based B1 mapping technique and we illustrate the effectiveness of this approach with in vivo B1 shim at 7Tesla.

Principle: This technique is proposed for Transceiver arrays (each element used as Transmit and Receive) and relies on the following observations: 1) at very high field the Transmit and Receive profiles of a coil element differ with twisted patterns following opposite rotational directions, 2) RF coil arrays are usually azimuthally distributed around the imaged target (head, torso), sharing with the latter multiple levels of symmetry in an axial plane. We observed empirically (in experiments and simulated data) that the sum of magnitude of all |B1+| profiles tend to resemble the sum of magnitude of all receive |B1-| profiles. Let us consider N transceiver coils (N=16 here) with $|B1_k+|$ and $|B1_j-|$ profiles with index k in transmit mode and j in receive mode. We acquire (as previously described [1] for relative phase B1+ mapping) a series of N GRE images, one channel transmitting at a time (k:1=>N), with signal sampled on all receive channels. In the small flip angle regime ($\alpha \leq 10°$) and assuming negligible T1 weight, the signal sampled from coil j when transmitting with coil k can be expressed as $|S_{k,j}| \approx M_{z0}\lambda|B1_j-|.|B1_k+|$, with M_{z0} proportional to proton density and λ a scalar. We ignore here T2* and T2. The sum of the magnitude of the complete series (16 Tx times 16 Rx=256 images) becomes:
$\Sigma_k[\Sigma_j|S_{k,j}|] \approx M_{z0}\Sigma_j\{|B1_j-|\cdot[\Sigma_k|B1_k+|]\} = M_{z0}\lambda[\Sigma j|B1_j-|][\Sigma_k|B1_k+|]$. By substituting $\Sigma_k|B1_k+|$ with $\Sigma j|B1_j-|$, we write $\Sigma_{k,j}|S_{j,k}| \approx M_{z0}\lambda[\Sigma k|B1_k+|]^2$. Finally, we obtain the following approximation: $[M_{z0}\lambda]^{1/2}\cdot\Sigma_k|B1_k+| \approx [\Sigma_{k,j}|S_{j,k}|]^{1/2}$, which estimates the sum of magnitude of all $|B1_k+|$, biased (multiplied) with the square root product of M_{z0} with λ. However, M_{z0} (proton density) typically varies by less than 20% through brain tissues, (whereas |B1+| can vary several fold), and even less in a root squared form. The relative contribution of each transmit coil is: $R_k = \Sigma_j|S_{k,j}| \div \Sigma_{j,k}|S_{k,j}|$. Finally, each $|B1_k+|$ is estimated with $|B1_k+|\cdot M_{0z}^{1/2} = R_k\cdot[\Sigma_{k,j}|S_{j,k}|]^{1/2} \div \lambda^{1/2}$. The scalar λ is invariant through space and can be estimated in a single location, using any standard power calibration.

Methods: We used a 7T scanner (Siemens) equipped with 16 x 1kW CPC RF amp (CPC™) with and an elliptical 16 channels head transceiver [2]. Volunteers signed an IRB approved consent form. The fast B1 calibration data, within an axial plane, consisted in a series of a 16 Gradient Echo (GRE) images one channel transmitting at a time in the small flip angle regime (nominal flip angle=10°), one image without RF pulse for noise analysis and one image with all channels transmitting for consistency check. (total acquisition time=1min10sec) We utilized the AFI technique [3] to obtain 3D B1 maps through the whole brain in 3min22sec (all coil transmitting together). The 16 *measured* |B1+| maps were derived from the small flip angle series *and* the 3D B1 map, based on B1 interferences as described in [4].

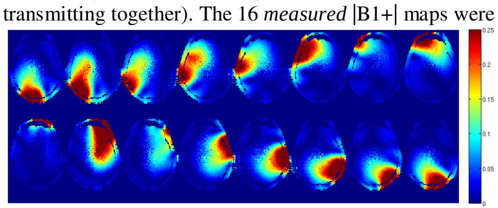

Fig 1A (relative transmit B1)

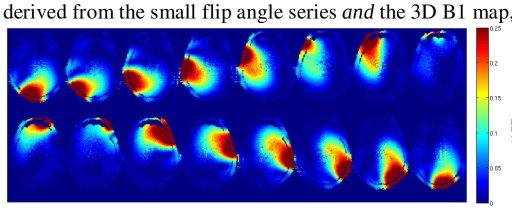

Fig1B (relative receive B1)

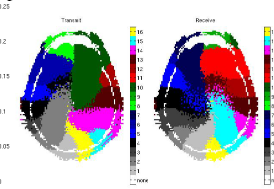

Fig1C

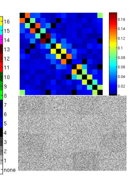

Fig1D

Results and Discussion: As shown in Fig.1, several parametric maps can be derived from the fast, low flip angle acquisition, such as relative Transmit B1 and relative Receive B1 (Fig 1A and 1B). For each pixel, we determine which Transmit coil and which Receive coil contributes the most to the signal as shown in Fig.1 C, where each color represents a coil. A noise correlation matrix and raw noise maps are shown in Fig1D. Those calibration results are of importance here because our B1 estimation algorithm assumes that transmit and receive sensitivities are evenly distributed through the coil elements and do not suffer from hardware related gain variations which could alter existing patterns of symmetry. In Fig.2 A are shown the 16 *estimated* Transmit |B1+| maps as derived solely from the small flip angle series based on the equations above [Principle section]. The gray picture (Fig.2B) shows the term $[\Sigma_{k,j}|S_{j,k}|]^{1/2}$ which, as expected, carries some Proton Density (PD) contrast (actual |B1+| maps do not carry tissue contrast). On the other hand, the pattern of signal intensity is overall smooth, with a brighter periphery and without evident visibility of the individual coil elements. It is interesting to note that this contrast is however much less visible in the estimated B1 maps than in Fig2B.

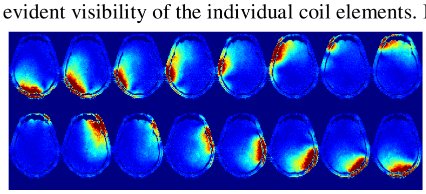

Fig 2A (estimated B1+)

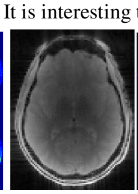

Fig2B

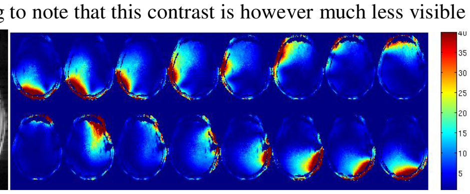

Fig2C (measured B1+ in flip angle equivalent)

Those estimated B1 maps should be compared to those in Fig.2C, obtained with a multichannel B1 mapping technique which does involve a large flip angle acquisition[1]. Even though these maps are not identical, they share very strong similarities and an interesting question is to determine whether those estimated maps could be successfully utilized in actual B1 applications such as B1 shim or Transmit SENSE. For this purpose, an ROI was drawn in the occipital lobe (white ROI Fig3) and two B1 Phase Shim solutions were calculated and applied in order to obtain an homogeneous B1 magnitude within this ROI: one based on the *estimated* B1 maps, another one based on the *measured* B1 maps. As can be seen in Fig3 (in degrees), both B1 Shim sets produced tremendous improvement in B1 efficiency and B1 homogeneity (the same RF power was used for the 3 AFI B1 maps, nominal flip angle 70°. Only B1 Phases were adjusted with B1 Shim). The mean flip angle [+/-std] rose from 40° [+/-6.9] in Fig4A (before B1Shim) to 69° [6.0] in Fig4B (B1 Shim based on measured B1 maps) and 73° [7.9] in Fig4C (B1 Shim based on estimated B1 maps).

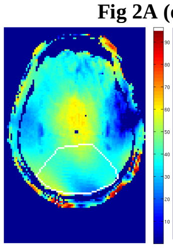

Fig3A

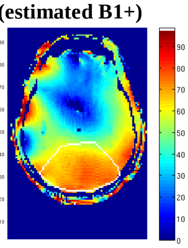

Fig3B

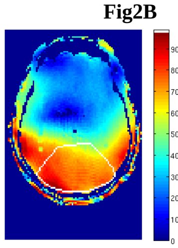

Fig3C

Conclusions and Discussion: We have described a fast method to estimate 16 transmit B1 profiles in a transceiver array coils in the small flip angle regime with about 1 minute of total data acquisition time. Our results suggest that, despite of expected residual biases, it is possible to obtain excellent B1 Shim results with calculations based on these estimated B1 maps. Further investigation will help determining if this fastt B1 estimation could become part of common scanner calibration routines, such as B0 mapping, for integrating transmit B1 adjustment in standard MR sesions at high field. These maps can easily cover the whole brain (in less than 3 minutes for 40 slices) and could also be used to determine a good B1 Shim set as a starting point for mapping B1 (with all coils transmitting) with more conventional techniques.

References: [1] Van de Moortele et al. ISMRM 2007 p1676, Metzger et al. MRM2008,59:396-409, [2] Adriany et al. MRM2008 [3] Yarnykh et al, MRM2007,57:192-200 [4] Van de Moortele et al. ISMRM2007, p1676 **Acknowledgments:** KECK Foundation. EB006835, PAR-02-010, EB007327, P41 RR008079 and P30 NS057091.