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

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**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.

**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+] and [B1-] profiles with index i 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 (α<10°) and assuming negligible T1 weight, the signal sampled from coil j when transmitting with coil k can be expressed as [S_k,j]=M_0[|B1_k|][|B1_j|], with M_0 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 Rxs=256 images) becomes: Σ_i(Σ_j[|B1_k||B1_j|]+Σ_j[|B1_k|])=M_0λΣ_j[|B1_j|]. By substituting Σ_j[B1_j] with Σ[B1_+], we write Σ_i[|B1_k|]=M_0λΣ_j[|B1_[+]|]. Finally, we obtain the following approximation: M_0λ=Σ_j[|B1_+|]/Σ_j[|S_j|]1/2, which estimates the sum of magnitude of all [B1+] biased (multiplied) with the square root product of M_0 with λ. However, M_0 (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=Σ_j[|S_j|]/Σ_j[|S_j|]. Finally, each [B1+] is estimated with [B1_+]=M_01/2=R_k[Σ_j[|S_j|]]1/2[|B1+|]. 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 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 interferes as described in [4].

**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 [Σ_j[|S_j|]]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 Fig.2B.

**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 estimated 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 fast B1 estimation could become part of common scanner calibration routines, such as B0 mapping, for integrating transmit B1 adjustment in standard MR sessions 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.