B_1 mapping based on signal modeling with magnetization prepared 3D EPI

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Introduction: Inhomogeneity of the excitation radiofrequency (RF) field (B_1) due to wavelength effects is a well known problem in human MRI. It can lead to intensity variations in the acquired images, affect the contrast in T_1-based anatomical imaging methods, and give rise to so-called "hot-spots", leading to possible patient safety issues. B_1 non-uniformity becomes more pronounced at higher field strengths, where the wavelength of the RF field starts to approach the size of the object. Several methods have been proposed for measuring the B_1 field distribution. Recently, it was suggested to measure the distribution based on the period of signal magnitude variation in a series of images acquired after a non-selective preparation pulse of varying flip angle [1,2] (figure 1). In this abstract, we present a modification of the method proposed in [2], which we combine with a 3D EPI acquisition. This allows rapid acquisition of B_1 field maps, and leads to a simple mathematical description that allows taking into account the effects of the imaging flip angle and the selected TR.

Theory & Methods: A 3D image acquisition scheme was implemented using single shot in-plane EPI acquisitions and an additional phase encoding gradient for the third (z) dimension. Zero-order navigator correction was used to account for phase errors between shots. Each imaging excitation was preceded by a sinc modulated preparation pulse played without selection gradient, and a spoiler gradient. Each 3D volume was acquired with 32 (z) phase encode steps. Since B_1 variations are expected to be low spatial frequency, low resolution matrices were elected. Each volume was imaged 15 times, and the nominal flip angle of the preparation pulse was stepped linearly between subsequent volume acquisitions, covering a range from 0° to 420° degrees. In order to avoid high SAR values, TR was set to 500ms at 3.0T and to 750ms at 7.0T leading to a total acquisition time of 4min and 6min respectively.

Assuming dynamic equilibrium during most of the 3D scan, it can be shown that the signal is proportional to $\cos(\lambda_{prep}\ \varphi_{prep})/[1-b*\cos(\lambda_{imag}\ \varphi_{imag})*\cos(\lambda_{prep}\ \varphi_{prep})]$ (1), where φ_{prep} and φ_{imag} are the nominal angles of the preparation and the imaging pulses, and a and b are constants dependent on the imaging flip angle, tissue T_1 and TR. Parameters λ_{prep} and λ_{imag} describe the ratio of the actual to the nominal flip angle, and they should be the same in most of the volume, except at the transition and out-of-band regions of the selective imaging pulse.

In order to produce a map of the excitation field, we minimized the least squares error between the rectified form of (1) and the modulus data using a downhill simplex method [3]. Since the cost function exhibits several local minima, in order to ensure convergence to the global minimum the fitting was repeated using 9 different sets of initial values, covering a range of λ_{prep} from 0.4 to 2. The range can be adjusted based on the expected values for λ_{prep}. The solution exhibiting the smallest error was selected.

Materials & Results: The proposed method was tested on a GE 3.0T scanner using a body coil for the transmission and a 16-channel array headcoil for the reception, and on a GE 7.0T scanner, using a birdcage coil for the excitation and an 8-channel array headcoil for the reception. Head coils were built by Nova Medical. The sequence was tested on phantoms and on normal volunteers, scanned after giving informed consent under an IRB approved protocol. Overall, there was good agreement between experimental data and fitted curves, independent of the selected imaging flip angle, as seen in figure 1. The multiple initial value sets provided good stability over a large range of λ_{prep}, as seen in a B_1 map from a spherical phantom acquired at 7.0T (figure 2), exhibiting a bullseye RF field distribution. Figure 3 shows the measured B_1 distribution in the head of a normal volunteer at 3.0 T, with refocusing of the RF in the center of the brain. Figure 4 shows the B_1 map from the head of a normal volunteer at 7.0T, where the shorter wavelength leads to a different distribution.

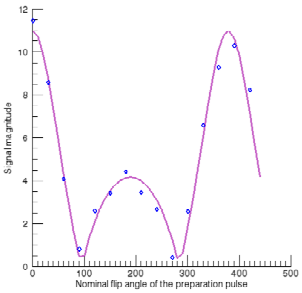

Figure 1: Experimental data (blue dots) and fitted model line (purple) for 60° nominal imaging angle.

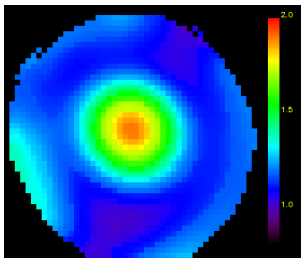

Figure 2: B_1 distribution in a spherical phantom, acquired at 7.0T

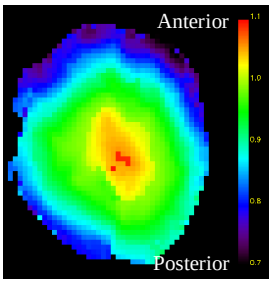

Figure 3: B_1 distribution in the head of a normal human volunteer (axial oblique slice), acquired at 3.0T

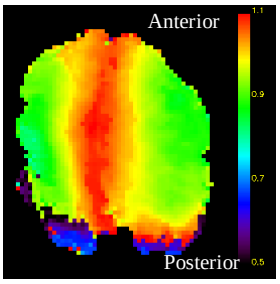

Figure 4: B_1 distribution in the head of a normal human volunteer (axial oblique slice), acquired at 7.0T. Shimming problems cause signal loss and poor fitting at the back of the brain.

Discussion & Conclusion: We have presented an EPI-based method for rapid acquisition of 3D maps of the excitation B_1. The technique shows sufficient stability over a wide range of λ_{prep}. Use of a three-dimensional imaging technique has the advantage of minimizing errors associated with the profile of the selective imaging pulse. At the same time it allows a simple mathematical expression of the expected signal behavior. The signal is not acquired at a relaxed state, but in dynamic equilibrium, eliminating the need for a relaxation time interval. Imaging angles less than 90° are acceptable, and possibly better for fitting stability. Since SNR is enhanced due to the 3D acquisition, shorter TR values can also be considered. Both the imaging flip angle and the TR can be subject to further optimization.

References [1] Vaughan JT et al., MRM 46:24-30 (2001), [2] De Vita E et al., Proc. 11th ISMRM, p2090 (2004), [3] Press WH et al., *Numerical Recipes in C: The Art of Scientific Computing, Second Edition*, Cambridge University Press (1992)