B1 Mapping with whole brain coverage in less than one minute

S. Volz[1], U. Nöth[1], and R. Deichmann[1]
[1]Brain Imaging Center (BIC), Goethe University Frankfurt, Frankfurt am Main, Germany

Introduction There is great demand for fast B1 mapping techniques, e.g. for correction of magnetization transfer ratio maps [1] or of quantitative T1 maps [2]. Several methods have been proposed, such as the acquisition of a spin echo and a stimulated echo (STE) [3], the performance of two acquisitions with different excitation angles [4], or the interleaved acquisition of two FLASH images with different repetition times TR [5]. The problem with most of these techniques is the relatively long duration, in general several minutes for whole brain coverage. Only one B1 mapping technique with subminute duration for whole brain coverage has been described so far, based on STE images with two different excitation angles [6]. The purpose of this study was the development of a fast B1 mapping technique based on magnetization prepared FLASH imaging. The use of specially designed selective preparation and excitation pulses allows for multislice imaging, yielding an acquisition time of 46 s for whole brain coverage with an in-plane resolution of 4 mm and a slice thickness of 3 mm.

Theory The method is based on the acquisition of two multislice FLASH data sets with excitation angle α_0, the second one being prepared by slice selective RF pulses that tilt the longitudinal magnetization by a nominal angle β_0, followed by a crusher gradient. Thus, the signal quotient of image intensities yields the cosine of the actual angle β, and B1 follows from β/β_0. There are mainly two conditions for the accuracy of this method:

1. The preparation angle must be constant across the excited slice. With $-1<u<1$, pulse shapes were given by $\mathrm{sinc}(n*\pi*u)*\cos(\pi/2*u)$; preparation pulse: n=8, duration P=5120 us, reference gradient (for a 1 cm slice) G=8.5 mT/m; excitation pulse: n=2, P=2560 us, G=7 mT/m. Numerical solution of the Bloch equations for $\beta_0=45°$ and $\alpha_0=30°$ shows that for the chosen pulses the deviation of β across the excited slice does not exceed 1%.

2. Image intensities must be proportional to the initial longitudinal magnetization, requiring centric phase encoding (PE) [7]. However, during the acquisition the magnetization still relaxes with a time constant T1* given by $1/T1^*=1/T1-1/TR*\log[\cos(\alpha_{eff})]$ with $\alpha_{eff}=0.87*\alpha$ [8], leading to different intensity quotients of peripheral k-space lines and thus potentially yielding erroneous B1 values for small structures. For correction, images are Fourier transformed in PE direction, and a k-space line acquired at time t after the preparation pulse is multiplied with the correction factor

$C(t)=K*[A+(K-A)*\exp(-t/T1^*)]^{-1}$ with: $A=T1^*/T1$. K is either 1 (reference image) or $\cos(\beta)$ (magnetization prepared image).

Corrected k-space data are subsequently transformed into image space. Basically, this correction requires the knowledge of local α and β values, but errors are small if the nominal values α_0 and β_0 are used. Another problem arises from the fact that the correction is T1 and thus tissue dependent. There are two methods of dealing with this issue: (1) global correction: an average brain tissue T1 of 1100 ms is assumed, as proposed in the literature [9]; (2) tissue dependent correction: data are segmented into white matter, grey matter and CSF subsets first, for each subset an individual k-space correction with the respective T1 value is performed, and the sum of corrected k-space data is transformed back into image space.

Materials and Methods Measurements were performed on six healthy subjects, using a 3T whole body MR scanner (8-channel head receive coil, whole body transmission). The B1 mapping technique was implemented with the following parameters: matrix=44x64, FoV=176x256mm^2, in-plane resolution 4 mm, 44 slices with 3 mm thickness, TR/TE/α_0=11ms/5ms/11°, β_0=45°, duration 46s. For reference, the method proposed in [5] was used with identical matrix size, FoV, and volume coverage, TR1/TR2/TE/α=30ms/100ms/3.13ms/60°, duration 5min 43s. B1 evaluation was performed in three ways: (1) no correction for relaxation effects, (2) global correction; (3) tissue dependent correction.

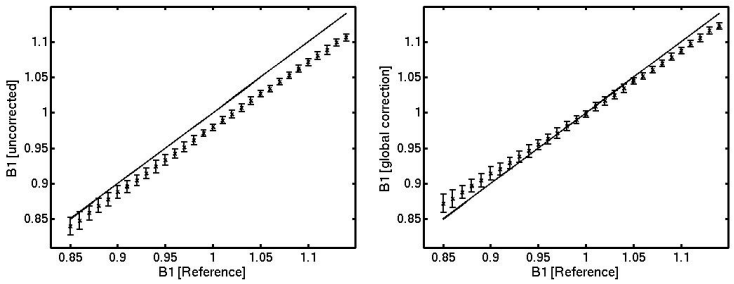

Fig 1: B1 values obtained without correction for relaxation effects (left) and with global correction assuming a T1 of 1100 ms (right). The straight line corresponds to equality of values measured with the proposed method and the reference values.

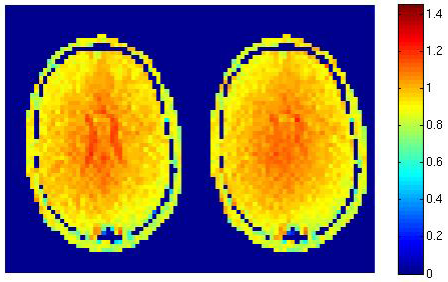

Fig 2: B1 maps obtained with correction of relaxation effects, using a global T1 of 1100 ms (left) and tissue dependent T1 values (right).

Results Figure 1 shows a comparison of reference B1 values and the results obtained with the proposed method, either without correction of relaxation effects (left), or with global correction (right). Even without correction, systematic errors are small, amounting to 3-4% for large B1. With global correction, errors are reduced below 2%. Residual errors are due to the fact that the nominal values for α and β are used for the correction. Figure 2 shows a single slice from the B1 map obtained for a single subject, with global correction (left) and with tissue dependent correction (right). Maps are almost identical, but for the global correction B1 values in the ventricles are systematically too low, obviously due to the prolonged T1 values of CSF that deviate considerably from the assumed value of 1100 ms. However, the errors in the ventricles amount to 3% only.

Discussion The method presented allows for fast B1 mapping with whole brain coverage, an in-plane resolution of 4 mm and a slice thickness of 3 mm within 46 s, with an accuracy of 2%. It is based on magnetization prepared FLASH imaging with centric PE. For improved accuracy, a correction for relaxation effects during the readout is required. Although correction factors depend on T1, global correction with the assumption of an average T1 value of 1100 ms is fully sufficient. Due to the use of short echo times and low excitation angles, the method hardly suffers from magnetic field distortions and has a low specific absorption rates (SAR), making it particularly attractive for the use at high magnetic fields.

References

[1] Samson et al., 2006. MRI 24, 255-263. [2] Preibisch and Deichmann, 2009. MRM 61, 125-135. [3] Akoka et al., 1993. MRI 11, 437-441. [4] Stollberger and Wach, 1996. MRM 35, 246-251. [5] Yarnykh, 2007. MRM 57, 192-200. [6] Helms et al., 2008. MRM 60, 739-743. [7] Vaughan et al., 2001. MRM 46, 24-30. [8] Deichmann, 2005. MRM 54, 20-27. [9] Jiru and Klose, 2006. MRM 56, 1375-1379.