Whole volume three dimensional B1 mapping in 10 seconds

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INTRODUCTION: Many methods for quantification of MR parameters, such as T_1 and T_2 relaxation and proton density would become far more accurate in case the exact amplitude of the radio-frequent B_1 field throughout the measured volume is included in the quantification fitting algorithm. For conventional clinical contrast imaging knowledge of the B_1 homogeneity could be used to rescale the image intensity for a uniform appearance. In most cases, however, B_1 is assumed to be equal to the desired field throughout the volume. The reason is that determination of the B_1 map is not so a straightforward procedure. Several methods for B1 quantification has been presented in the literature, however, the most important limiting factor preventing most methods to be used in clinical protocols is very long scan times.

A method is presented that can measure the absolute B1 field of a complete volume in a mere 10 seconds. It is largely independent of pulse profile, of coil sensitivity, of relaxation parameters (T1, T2) and of the total signal strength (proton density and RF amplification). Moreover, it calculates B1 field on the basis of real images, such that there is no ambiguity of the sign of the signal and noise distribution in Gaussian rather than Rician. The latter is very important in areas with low signal strength. For the sequence no scanner software changes were required, a standard TFE sequence using a saturation pre-pulse was easily adapted for the measurement.

THEORY: The complete quantification measurement consists of a series of low flip angle shots at increasing delay time T_D after a saturation pulse θ were a flip angle α is applied every repetition time T_R. Due to the continuous application of small flip angles the apparent relaxation time, T_1*, is shorter than the actual longitudinal relaxation T_1 and the magnetization approaches a saturated magnetization, M_0*, lower than the unsaturated magnetization M_0, (Eq. 1).

The magnetization M_{TD} at a specific delay time T_D after the saturation pulse θ can be found using the recursive relation of magnetization over the total cycle time, T_C, between consecutive saturation pulses, (Eq. 2).

From the T_1* relaxation curve the local saturation flip angle, θ_{eff}, can be found and hence the local B_1 field. This is done using the ratio of M_{T0} at time 0, just after the saturation pulse and the magnetization M_{TC} at time T_C, just before the subsequent saturation pulse, (Eq. 3), since the difference between M_{T0} and M_{TR} is entirely due to the saturation pulse and subsequent spoiling.

$$\frac{M_0^*}{M_0} = \frac{T_1^*}{T_1} = \frac{T_R}{T_R - T_1 \ln(\cos\alpha)} \quad (1)$$

$$M_{TD} = M_0 \frac{1-(1-\cos\theta)\exp\left(-T_D / T_1^*\right) - \cos\theta \exp\left(-T_C / T_1^*\right)}{1-\cos\theta\exp\left(-T_C / T_1^*\right)} \quad (2)$$

$$\theta_{eff} = \operatorname{acos}(M_{T0} / M_{TR}) \quad (3)$$

MATERIALS AND METHODS: All experiments were performed on a 1.5T Achieva scanner (Philips Medical Systems, Best, the Netherlands). The method consists of an adjusted TFEPI sequence with a saturation pre-pulse followed by a number of ('cardiac') phases. The saturation pre-pulse is non-slice-selective and shared for all phases. For triggering the physiology simulation function was used. A quadrature receive/transmit head coil with a field of view of 210 x 180 mm with 3 mm resolution, slice thickness 5 mm was used for acquisition of brain images of a healthy volunteer. The whole volume consisted of 27 slices acquired in a scan time of 10 s. There were 4 phases over 2 s cycle time using a TFE factor of 60, the EPI factor was 9. The profile order was chosen low-high in order to measure the signal intensity as close after the saturation pulse as possible.

The acquired data was analyzed as follows: The phase of the latest image after the saturation pulse was used as a reference phase. This phase was subtracted for all other images to generate real images. The real data was fitted with a bi-exponential function in order to account for possible partial volume effects. The fitted curve was extrapolated to time zero just after the saturation pulse and to time T_C just before the subsequent saturation pulse. The saturation flip angle θ_{eff} was calculated according to Eq. 3. Since the B_1 field was assumed not to change rapidly over the volume θ_{eff} was smoothed over 10 mm using a median filter. Furthermore, θ_{eff}, was assumed to be linearly dependent on the B1 field strength.

For validation the proposed method is compared to the B1 estimation of a flip angle sweep method. The flip angle α was set to the range [30 50 70 90 110 130 150] degrees, the TR was 10 s. The B1 field was calculated using a least square fitting routine to fit the intensity $S \propto \sin(c \cdot B_1 \cdot \alpha)$.

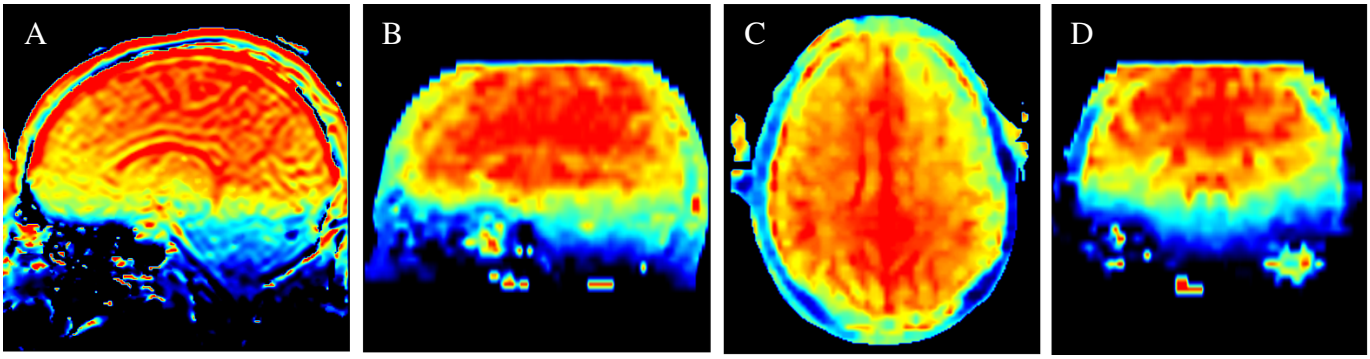

Fig 1. A) A B1 field map generated using the reference flip angle sweep method (total scan time 14 minutes). B-D) Three orthogonal views of B1, acquired using the presented method (total scan time 10 seconds). The colormap used was the same in all four images.

RESULTS: In Fig 1A a sagital B1 field map, acquired using the reference flip angle sweep method in 14 minutes is shown. Note the residual structures in areas with CSF, caused by T1 saturation effects. In Figs B-D B1 field map images in three different orthogonal planes are displayed, acquired using the presented method in only 10 seconds. A high geometrical correlation was achieved between the two B1 field maps.

DISCUSSION: A method capable of producing 3D whole volume B1 field maps in just 10 s is reported. The method is flexible and can easily be adjusted to fit different organs. Furthermore it was implemented on a clinical MR-scanner without any pulse programming. The method needs further validation to determine the dynamic range and sensitivity. Stimulated echoes may cause artifacts when the effective saturation pulse angle deviates much from pi/2 or when compounds characterized by extremely long T2 (i.e. CSF) are imaged. However, as the method also gives an estimate of T1 in the voxels CSF is easily identified and its effect on the B1 field map may be canceled using more advanced filtering techniques than those used presently.