B1 INHOMOGENEITY CORRECTED T1-QUANTIFICATION FOR DGEMRIC USING 3D LOOK-LOCKER TECHNIQUE WITH NON-SLICE SELECTIVE RF-PULSES

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Introduction

3D Look-Locker (LL) delayed Gadolinium enhanced MRI of cartilage (dGEMRIC) is a technique for molecular imaging of the proteoglycan level in cartilage using quantitative T1 measurements [1]. However, such 3D T1 quantifications are typically sensitive to flip angle (FA) variations, caused both by B1 inhomogeneities and slice dependent FA variations due to the pulse shape. In this work a method is introduced to eliminate all such effects by using the same type of non-slice selective pulses for both excitation and inversion, in combination with an improved evaluation method.

Methods

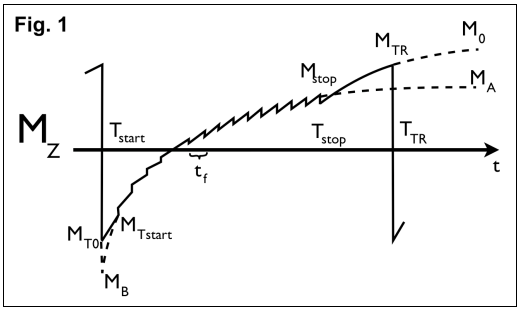
In the Look-Locker technique, the MR signal is sampled using a train of small FA RF pulses following an inversion pulse. A pseudo-T1 (T1*) relaxation time is then calculated, together with variables M_A and M_B, using a three-parameter fit to the LL data. Traditionally, the true T1 is then calculated from equation 3 [1], using only the nominal FA from the user interface (constant FA correction), with possibly erroneous T1 values as a result. It has previously been shown [2], that equations 1 and 2 will solve the true T1 for any LL data, given that the quality of the inversion (0<K<1) is known. For in vivo measurements however, K usually unknown since it fluctuates slightly spatially, resulting in an erroneous calculation of T1. By using a non-slice selective rectangular RF pulse for both excitation and inversion (with all pulse parameters equal except for the amplitudes), the two pulses will suffer from exactly the same B1 inhomogeneity effects, thus establishing the relation in equation 4. By combining equations 1-4 for each voxel and solving iteratively it is possible to calculate the correct T1 values (full B1 correction).

Eq. 1
$$T_1 = -\frac{t_f}{\ln\left(1 - \frac{M_A - M_A \cdot e^{-\frac{t_f}{T_1^*}}}{M_0}\right)}$$

Eq. 2
$$M_0 = \frac{M_{Tstart} + M_{Tstop} \cdot K \cdot e^{-\frac{T_{start}}{T_1}} \cdot e^{-\frac{T_R - T_{stop}}{T_1}}}{1 - e^{-\frac{T_{start}}{T_1}} - K \cdot e^{-\frac{T_{start}}{T_1}} + K \cdot e^{-\frac{T_{start}}{T_1}} \cdot e^{-\frac{T_R - T_{stop}}{T_1}}}$$

Eq. 3
$$\frac{1}{T_1} = \frac{1}{T_1^*} + \frac{\ln(\cos(\alpha))}{t_f}$$

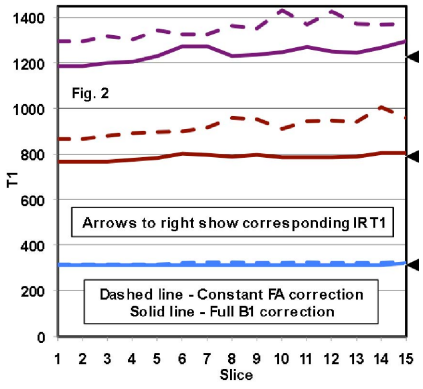
Eq. 4
$$K = \frac{\alpha}{\alpha_{Nom}}$$

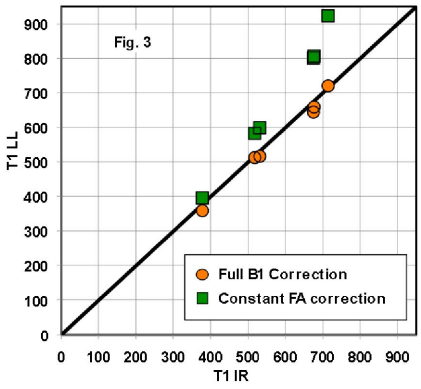


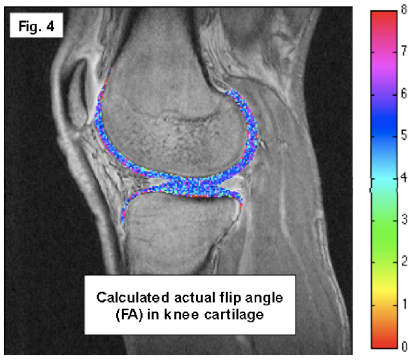
All imaging was performed on a Siemens Magnetom Sonata 1.5 T scanner and a CP Extremity coil. Three gel-filled tubes with different T1s were imaged with an in-house developed 3D LL (FOV = 16^2 cm^2, Matrix = 256^2, 30 slices, TR 2500 ms, FA = 6°, 11 contrasts, non-slice selective RF pulses). The tubes were aligned to the slice-encoding direction, such that the tubes covered slice 1-15 of the acquired slices. T1 was calculated for those slices using both constant FA correction and full B1 correction. The same sequence was also used for in vivo imaging of 3 subjects (sagittal orientation, 3D volume centered at the knee midline between the condyles). Sequence duration was 10 min 42 s. Standard 2D IR dGEMRIC was also acquired as a reference. T1 was evaluated with ROIs in mid-lateral and mid-medial condyle slices.

Results

Due to B1 inhomogeneities, T1 calculated using traditional constant FA correction is generally overestimated in all phantoms (fig 2, Phantom measurements), especially for longer T1. This effect is also prominent in vivo (fig 3). If this is not compensated for, the constant FA correction is likely to fail in restoring the T1 obtained by reference IR measurements. The full B1 correction however, successfully corrects for all B1 inhomogeneities (fig 4). With this method the measured T1 is accurate throughout all slices in the phantoms (fig 2) and also throughout all in vivo measurements (fig 3).







Discussion and conclusions

By using non-slice selective pulses the effective FA is fairly constant within all slices, due to the absence of a pulse slice profile, which is beneficial for T1 measurements. However, the non-slice selective pulses also results in folding effects if the investigated object extends outside of the covered slices. In knee-dGEMRIC this is not a problem, since the width of a knee is within the acceptable slice volume coverage of the sequence. In other applications of 3D LL where the investigated objects are larger, such as hip-dGEMRIC, this method might not be as suitable.

References

1. Kimelman et al, Inv. Radiol. 41:198-203 (2006), 2. Siversson et al, ISMRM No.5289 (2008)