Accurate and Rapid T1 Measurements for First-Pass Quantitative Perfusion Studies

A. Cernicanu[1], D. Kim[2], V. Pai[2], T. Manglik[2], L. Axel[2]

[1]Electrical and Systems Engineering, University of Pennsylvania, Philadelphia, PA, United States, [2]Department of Radiology, New York University, New York, NY, United States

Introduction

Quantitative perfusion measurements could potentially detect the presence and extent of ischemia, with significant impact on clinical decisions related to re-vascularization and re-perfusion. However, methods for quantitative MRI first-pass perfusion studies are still under development [1]. A typical approach in MR cardiac perfusion imaging is to acquire T1-weighted images at several short-axis locations in every (or every other) cardiac cycle during the first pass of a bolus of contrast agent. However, the conversion of T1-weighted image signal to tracer (Gadolinium chelate) concentration ([Gd]) using in vitro phantom-based calibration data has limitations for in vivo studies, since it uses a fixed set of imaging parameters. We have developed a method for accurately measuring T1 during first-pass imaging, which enables us to convert the image signal to [Gd] without the need for empirical in vitro calibration.

Materials and Methods

We acquired serial T1-weighted images with a modified fast gradient echo - echo planar imaging (FGRE-EPI) pulse sequence [2] with an interleaved center-out k-space acquisition. Considering longitudinal magnetization evolution only, we derived a theoretical expression that relates the signal (S) of the T1-weighted images obtained with such a sequence to T1 and to imaging parameters. T1 weighting is obtained by playing a 90° saturation pulse and allowing partial recovery (during a delay, TD) before the read-out. The signal S is directly proportional to the longitudinal magnetization value at the time when the center of k-space is acquired, and therefore it is not the result of a simple exponential recovery during TD, but of a more complex expression that includes the flip angles, TR, and the number of TR intervals before the k-space center acquisition. This expression can be used to map a single signal value onto a T1 value. Thus, what we propose is a single-point T1 measurement method, conditioned by: **1)** the correctness of the S(T1) expression and negligible T2 dependence; **2)** complete signal saturation at the beginning of each image acquisition (i.e., ensuring that the T1 recovery starts from zero) and **3)** elimination of the receive coil inhomogeneity profile and the unknown equilibrium longitudinal magnetization from S(T1) through normalization by a corresponding proton density (PD)-weighted image (i.e., fixing the normalized asymptote for the T1 exponential recovery curve to 1). If these conditions are met, the only limitation on the accuracy of the T1 measurement is the T1-weighted image SNR.

Our sequence acquires one PD-weighted image for each short-axis slice location in the first heartbeat interval, and then proceeds to the acquisition of a time series of T1-weighted images for the same slice locations (simulated ECG gating was used for phantom studies). We use a low imaging flip angle (8°), chosen based on a full Bloch equation simulation of magnetization evolution, in order to minimize k-space modulation-related artifacts and transverse coherences and thus maintain the validity of the S(T1) relationship (condition 1 above is met). The loss in SNR due to a smaller imaging flip angle is partially compensated by the fact that there is less interference with the longitudinal magnetization recovery from the train of imaging flip angles as k-space is traversed. The SNR reduction due to lower imaging flip angles was quantified in a uniform phantom with T1 = 280 ms. We use a B_1-insensitive rotation pulse with 4 adiabatic segments (BIR-4) to ensure complete saturation (condition 2 above is met). We imaged a calibration phantom consisting of spheres filled with Gd contrast agent (Magnevist, Schering AG) dilutions, with the following Gd concentrations: 0.00, 0.05, 0.10, 0.15, 0.20, 0.25, 0.30, 0.50, 1.0, 2.0, 3.0 and 6 mmol/L, and a normal volunteer (2 short axis slices). Reference phantom T1 measurements were obtained using a series of saturation recovery images with different recovery delays and a singe-shot EPI read-out. The imaging was performed on a 1.5-T MR system (Sonata, Siemens), using a phased-array surface coil. Imaging parameters included: TD = 75 ms, TR = 5.8 ms, echo train length of 4, acquisition matrix 128 x 96, FOV = 302 x 226 mm, in-plane resolution = 2.4 x 2.4 mm/pixel, slice thickness = 10 mm, receiver bandwidth = 1954 Hz/pixel. The initial PD-weighted image acquisition is acquired without a saturation pulse, and the PD-weighted imaging flip angle is reduced to 4° in order to minimize T1 weighting. When processing the images (using Matlab [The MathWorks, Natick, MA]), for each slice, all images in the time series of T1-weighted images are divided on a pixel-by-pixel basis by a smoothed and scaled version of the initial PD-weighted image (condition 3 above is met). Using S(T1), the normalized image signal intensities are then mapped to T1 values for both in vitro and in vivo images.

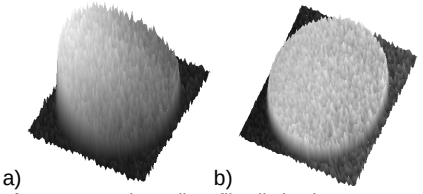

Figure 1. Receive coil profile elimination (a) Original and (b) normalized T1-weighted image

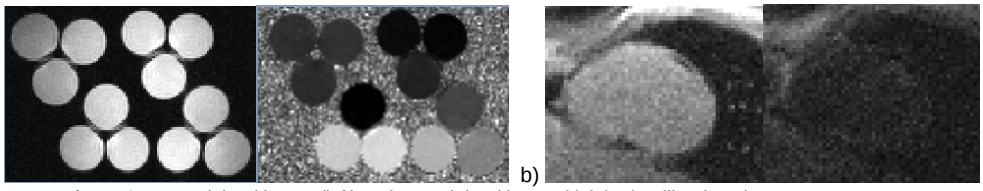

Figure 2. PD-weighted image (left) and T1-weighted image (right): a) calibration phantom; b) mid-ventricular short-axis slice.

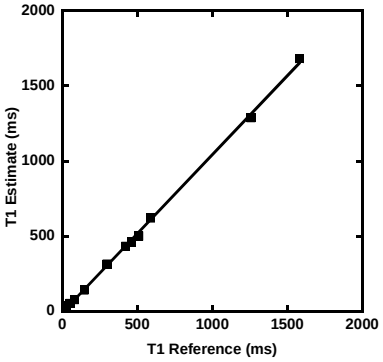

Figure 3. T1 measurements correlation for 11 different [Gd] values

Results

Fig. 1 demonstrates the elimination of the effects of the receive coil inhomogeneity profile in a large uniform phantom. Fig. 2 shows representative PD-weighted and T1-weighted images of the multi-concentration calibration phantom and of the heart. The images acquired with a low imaging flip angle have minimal artifacts, and The SNR loss sustained when reducing the imaging flip angle from 16° to 8° is approximately 1.5, and not 2. The comparison between in vitro T1 estimates obtained with our new rapid method and reference T1 measurements is shown in Fig. 3. The T1 measurements are strongly correlated (R = 0.99, slope = 1.05 and bias = -6.0). Representative T1 values measured in vivo are 1060 ms, 821 ms, and 232 ms for blood, myocardium, and fat, respectively.

Discussion

The T1 values estimated from single normalized T1-weighted images (single-point measurements) with our theoretical absolute signal calibration correlate well with the reference measurements (multi-point measurements). We conclude that this method should be useful for first-pass in vivo cardiac perfusion studies, since it can provide rapid and accurate serial T1 measurements, which can then be converted to absolute [Gd].

References

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