Available online at www.sciencedirect.com
ScienceDirect
ELSEVIER
Magnetic Resonance Imaging 25 (2007) 748–753
MAGNETIC RESONANCE IMAGING

T_2* measurements in human brain at 1.5, 3 and 7 T

Andrew M. Peters[a], Matthew J. Brookes[a], Frank G. Hoogenraad[b], Penny A. Gowland[a], Susan T. Francis[a], Peter G. Morris[a], Richard Bowtell[a,*]

[a]*Sir Peter Mansfield Magnetic Resonance Center, School of Physics and Astronomy, University of Nottingham, NG7 2RD Nottingham, UK*
[b]*MR Clinical Science, Philips Medical Systems, 5680 DA Best, The Netherlands*

Accepted 11 January 2007

Abstract

Measurements have been carried out in six subjects at magnetic fields of 1.5, 3 and 7 T, with the aim of characterizing the variation of T_2* with field strength in human brain. Accurate measurement of T_2* in the presence of macroscopic magnetic field inhomogeneity is problematic due to signal decay resulting from through-slice dephasing. The approach employed here allowed the signal decay due to through-slice dephasing to be characterized and removed from data, thus facilitating an accurate measurement of T_2* even at ultrahigh field. Using double inversion recovery turbo spin-echo images for tissue classification, an analysis of T_2* relaxation times in cortical grey matter and white matter was carried out, along with an evaluation of the variation of T_2* with field strength in the caudate nucleus and putamen. The results show an approximately linear increase in relaxation rate R_2* with field strength for all tissues, leading to a greater range of relaxation times across tissue types at 7 T that can be exploited in high-resolution T_2*-weighted imaging.
© 2007 Elsevier Inc. All rights reserved.

Keywords: T_2*; Human brain; Mapping; Field strength

1. Introduction

The recent availability of ultra-high field magnets operating at fields of 7 T and above for human imaging [1] requires reoptimization of pulse sequences and imaging protocols for operation at elevated field [2]. This optimization can only be carried out efficiently with an accurate knowledge of the relaxation times of the tissues involved. Accurate knowledge of T_2* relaxation times in brain tissue is important, in general, for setting the optimal echo time in gradient echo sequences and, in particular, for choosing the echo time that maximizes blood-oxygenation-level-dependent contrast in functional magnetic resonance imaging (fMRI) experiments [3,4]. In addition to providing useful information for pulse sequence and fMRI paradigm optimization, there is considerable interest in using relaxation time measurements to evaluate local iron content in human brain [5–8]. Such measurements are valuable because the state and the concentration of iron, particularly in deep grey matter structures, have both been shown to vary with age and to be changed in neurodegenerative disorders, such as Parkinson's disease and Alzheimer's disease [5–8]. T_2* shows particular sensitivity to iron content, which is enhanced at increased magnetic field strength.

Here we describe the measurement of the T_2* value of different tissues in the normal human brain at 7 T and the comparison of these measurements with results obtained at 3 and 1.5 T on the same six volunteers. Accurate measurement of T_2* in the presence of macroscopic magnetic field inhomogeneity is difficult because of enhanced signal decay due to intravoxel dephasing [9,10], which leads to underestimation of T_2*. The approach employed here [10] allowed the signal decay due to through-slice dephasing to be measured and removed from data, thus facilitating precise measurement of T_2* even at ultra-high field.

2. Methods

Scanning was carried out on Philips Achieva systems operating at 7, 3 and 1.5 T. T_2* maps were produced using the modified echo planar imaging (EPI) sequence shown in Fig. 1B. Here, phase-encoding gradient blips shown in the conventional EPI sequence of Fig. 1A are replaced with a phase-encoding gradient pulse applied immediately following slice selection, and the sequence is repeated n times with different phase-encoding gradient strengths so as to span k-space. The resulting multishot multiecho acquisition

* Corresponding author. Tel.: +44 115 9514737.
E-mail address: richard.bowtell@nottingham.ac.uk (R. Bowtell).

0730-725X/$ – see front matter © 2007 Elsevier Inc. All rights reserved.
doi:10.1016/j.mri.2007.02.014

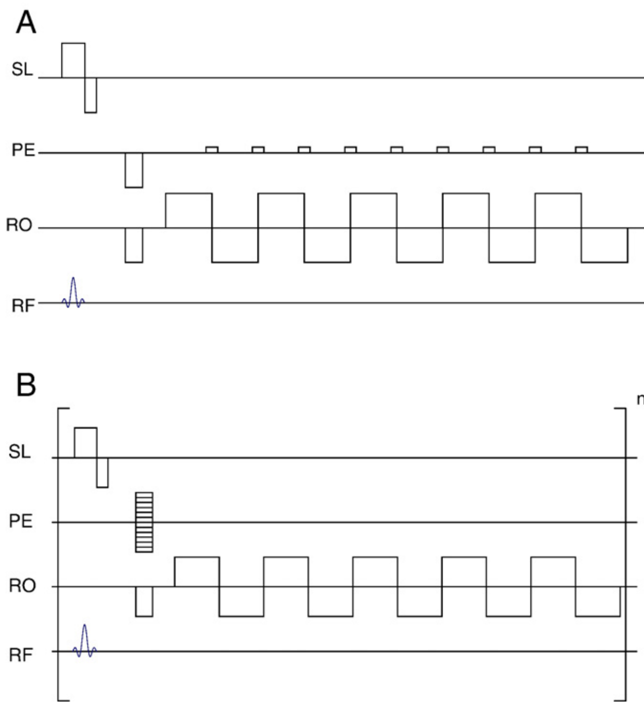

Fig. 1. (A) Timing diagram for a conventional single-shot EPI sequence, in which echoes are progressively phase encoded by application of a "blipped" phase-encoding gradient. (B) Timing diagram for the modified EPI sequence, which was used for T_2* mapping. Gradient blips are replaced by a conventional phase-encoding scheme.

allows the generation of one image from each of the echoes formed under the read-out gradient lobes, yielding a set of gradient echo images spanning a range of echo times.

Six subjects with a mean age of 37±11 years were scanned using identical protocols at each field strength. Two data sets, each comprising 16 slices with a 128×128 matrix size, were acquired at 64 echo times with a 270-mm field of view (FOV) and a 3-mm slice thickness in 2 min. T_R was set to 0.7 s, and a flip angle of 30° was used. The 64 echoes spanned a 45-ms range of echo times, and images formed from both positive and negative going-readout gradient periods were used in calculating T_2*. The first set of 16 slices was positioned to cover cortical grey matter and white matter, while the second was positioned so as to span the putamen and caudate nucleus. Grey-matter-selective and white-matter-selective images, such as those shown in Fig. 2, were acquired with the same resolution and matrix size as T_2* data using double inversion recovery (DIR) [11] in combination with a turbo spin-echo (TSE) sequence, so as to aid tissue classification in the analysis of T_2* data. Inversion times were chosen to null signals from the cerebrospinal fluid (CSF) and either grey matter or white matter at each field strength.

Signal loss due to dephasing caused by through-slice field gradients can lead to underestimation of T_2* since the signal decays more rapidly than expected. We therefore employed an approach that was recently proposed by Dahnke and Schaeffter [10] to correct for this signal decay. This approach assumes that the dominant effect of field inhomogeneity can be modeled as that of a constant field gradient, of strength G, in the slice direction, which causes the modulation of exponential signal decay by a sinc function. The variation of signal amplitude S with echo time T_E is then characterized by

$$S(T_E) = S_0 \exp\left(\frac{-T_E}{T_2*}\right) \operatorname{sinc}\left(\frac{\gamma G \Delta z T_E}{2}\right) \tag{1}$$

where Δz is the slice thickness [9] and S_0 is the signal strength at T_E=0. T_2* maps were obtained by pixelwise fitting of the measured signal to the expression shown in Eq. (1). The fitting process uses an initial estimate of $\gamma G \Delta z$, which is obtained from frequency maps produced by an evaluation of the linear rate of change of the signal phase with echo time. Uncorrected T_2* maps were also produced by applying a simple exponential fit to experimental data.

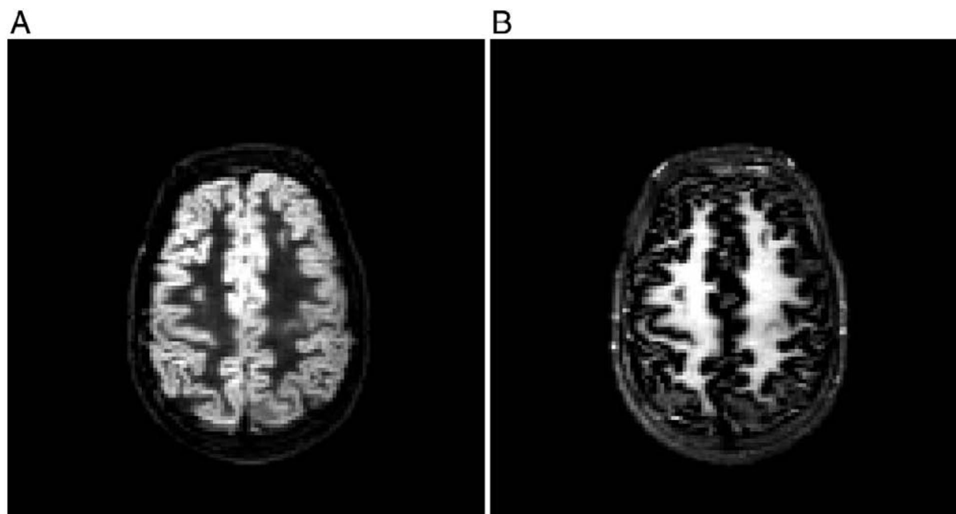

Fig. 2. DIR-TSE images acquired at 3 T with inversion times chosen so as to null the signal from (A) the white matter and CSF, thus yielding a grey-matter-selective image; and (B) the grey matter and CSF, thus yielding a white-matter-selective image.

Download English Version:

https://daneshyari.com/en/article/1807514

Download Persian Version:

https://daneshyari.com/article/1807514

Daneshyari.com