0004

Low-Angle Combined-Echo (LACE) Imaging in Highly Inhomogeneous B_0 Magnetic Fields

Sebastian Theilenberg[1], Chathura Kumaragamage[2], Scott McIntyre[2], Terry W. Nixon[2], Christoph Juchem[1,3], and Robin A. de Graaf[2]
[1]*Biomedical Engineering, Columbia University, New York, NY, United States,* [2]*Department of Radiology and Biomedical Imaging, Magnetic Resonance Research Center, Yale University School of Medicine, New Haven, CT, United States,* [3]*Radiology, Columbia University Medical Center, New York, NY, United States*

Synopsis

In line with recent developments in scanner design for more cost effective and more accessible scanners, we propose a low-angle combined-echo sequence similar to a conventional spin echo sequence capable of producing high quality images in the presence of strong B_0 inhomogeneity. We present simulation results investigating the signal dependence on the sequence's timings and flip angles as well as image contrast for typical relaxation times of normal brain white and gray matter. The simulations were validated in vivo at 4 T. Lastly, we show the feasibility to utilize multi-coil generated image encoding fields for this sequence.

Introduction

In recent years, a variety of new approaches to scanner design have been presented aiming to lower the costs of MRI and to increase accessibility. One of these approaches is relaxing the homogeneity requirements for the main B_0 magnetic field from typically less than a few hundred hertz to 20 kHz (equivalent to a few ppm to 300 ppm at 1.5 T) or more across the volume of interest in a recently proposed 1.5 T head-only scanner to reduce scanner cost[1]. This development necessitates the design of MRI pulse sequences robust against these conditions. Here we present a low-angle combined-echo (LACE) sequence capable of producing high quality images in the presence of strong B_0 inhomogeneity.

Methods

The proposed LACE sequence is, similar to a conventional spin echo sequence, based on two broadband radio-frequency pulses separated by the time $T_E/2$, signal readout at the time of the echo T_E and a delay to complete the repetition time T_R. With the use of short T_R delays, the echo formed at steady-state contains both spin echo and stimulated echoes (Figure 1). Simulations based on the product operator formalism[2] were performed in MATLAB (MathWorks, Natick, MA, USA) to investigate the dependence of the signal strength on the sequence parameters. The transverse magnetization was simulated at flip angles $\alpha_1/\alpha_2 \leq 40^\circ$, $T_E \leq 100\,\mathrm{ms}$ and $T_R \leq 150\,\mathrm{ms}$. The simulations have been performed with two combinations of T_1 and T_2 values ($T_1/T_2 = 900/50\,\mathrm{ms}$ and $1400/60\,\mathrm{ms}$) representing experimentally determined values for human brain white and gray matter (WM/GM) at 4 T.

To validate the simulation results, in vivo brain images were acquired from 6 healthy subjects on a 4 T Bruker scanner at the Yale MR Research Center (MRRC) with the following sequence parameters: $T_E = 12\,\mathrm{ms}$, $T_R = 100\,\mathrm{ms}$, AFPST4 pulse excitation (1 ms) with 20 kHz bandwidth, readout bandwidth BW_{RO} = 100 kHz. A paper clip was fixed to the bottom of the head rest below the subjects' head to validate the B_0 insensitivity of the MRI sequence at hand. The proposed 1.5 T head-only scanner will be equipped with a multi-coil (MC) array[3] instead of linear gradient coils for the generation of image encoding fields[1,4], therefore, we also tested LACE with MC generated encoding fields similar to previous work[5].

Results

The simulations show the transverse magnetization rapidly increases in the first few repetitions of the sequence and after reaching a maximum slowly decreases to a steady-state (Figure 2A). The steady-state magnetization is larger if α_2 is larger than α_1, and in general combinations of larger flip angles lead to a stronger steady-state magnetization, with a maximum in the investigated range around $\alpha_1/\alpha_2 = 30^\circ/40^\circ$ (Figure 2B). Steady-state magnetization increases with longer T_R and shorter T_E (Figure 2C).

Absolute contrast between GM and WM, calculated as the absolute difference between the steady-state transverse magnetization of the two sets of T_1/T_2 values, was found to be stronger with shorter T_E and longer T_R in general. There is an area of low contrast for intermediate values of T_R whose location is dependent on the relative proton density PD_{GM}/PD_{WM} between the two tissues (Figure 3A-C). A region of high contrast is evident when mapping it against the flip angles α_1/α_2. While the shape and strength of that area differ with relative PD, small flip angles around $\alpha_1 = 10^\circ$ and $\alpha_2 = 15^\circ$ should always yield good contrast (Figure 3D-F).

Brain images acquired in vivo match the simulated contrast behavior when assuming a relative PD of about 85% with images at $\alpha_1/\alpha_2 = 10^\circ/20^\circ$ showing good contrast with $T_E/T_R = 12/100\,\mathrm{ms}$, but images acquired at $\alpha_1/\alpha_2 = 10^\circ/40^\circ$ or with $T_R = 50\,\mathrm{ms}$ showing only little contrast (Figure 4). The latter case also shows reduced signal intensity, in line with Figure 2C. While the B_0 inhomogeneity induced by the paper clip effects large portions of a 3D gradient echo scan acquired for reference, LACE imaging is able to recover the signal in this area. LACE images acquired with MC generated encoding fields show largely the same image quality than the ones acquired with regular gradients (Figure 5). Mismatches between regular LACE and MC-LACE images are likely due to non-linearities in the MC encoding fields at the edge of the FoV.

Discussion

LACE MRI enables imaging in the presence of strong B_0 inhomogeneity by utilizing high bandwidth RF pulses and selecting combined spin- and stimulated echoes. The overall signal intensity is ~20% of a low-angle gradient-echo sequence (e.g. FLASH, calculated with the Ernst equation[6] at flip angle 15°) and is comparable to that of a recently described MP-SSFP method[7]. High GM/WM image contrast can be achieved with modest nutation angles, providing a low-power alternative to high-powered spin-echo-based methods (e.g. 3D FSE). Even though employing larger flip angles generates more signal, sufficient signal can be acquired at low flip angles with good GM/WM contrast. If the background B_0 field behavior is known, as it is the case for a given main magnetic field, the resulting image distortions can be corrected. Utilizing lower flip angles enables larger bandwidth RF pulses that are necessary to excite all spins in the presence of large off-resonances without exceeding SAR constraints. With only two excitation pulses the described sequence is rather simple and should be straightforward to implement in any MRI system.

Acknowledgements

This research was supported by the National Institute of Biomedical Imaging & Bioengineering of the National Institutes of Health under award number U01EB025153.

References

1. Garwood M, Mullen M, Kobayashi N, et al. A compact vertical 1.5T human head scanner with shoulders outside the bore and window for studying motor coordination. Proc Intl Soc Magn Reson Med. 2020;28:1278.

2. Sørensen OW, Eich GW, Levitt MH, Bodenhausen G, Ernst RR. Product operator formalism for the description of NMR pulse experiments. Prog Nucl Magn Reson Spectrosc. 1984;16:163-192.

3. Juchem C, Nixon TW, McIntyre S, Rothman DL, de Graaf RA. Magnetic field modeling with a set of individual localized coils. J Magn Reson. 2010;204(2):281-289.

4. Theilenberg S, Shang Y, Kobayashi N, Parkinson B, Juchem C. Multi-Coil Array for Combined Imaging and B0 Shimming in a Portable Head-Only Scanner. Proc Intl Soc Magn Reson Med. 2019;27:1480.

5. Juchem C, Theilenberg S, Kumaragamage C, et al. Dynamic multicoil technique (DYNAMITE) MRI on human brain. Magn Reson Med. 2020;6(84):2953-2963.

6. Ernst RR, Bodenhausen G, Wokaun A. Principles of Nuclear Magnetic Resonance in One and Two Dimensions. Oxford: Clarendon Press; 1987:610.

7. Kobayashi N, Parkinson B, Idiyatullin D, et al. Development and validation of 3D MP-SSFP to enable MRI in inhomogeneous magnetic fields. Magn Reson Med. 2020;85(2):831-844.

Figures

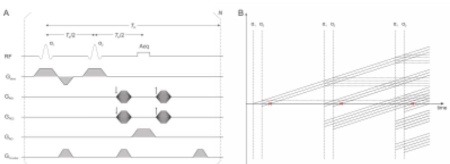

Figure 1: LACE sequence. A) Sequence diagram. B) Phase graph. The sequence consists of two broadband excitation pulses separated by $T_E/2$. Signal is acquired at the time of the spin echo of the two RF pulses (red crosses in B), which in the steady-state comprises a combination of signal from spin and stimulated echoes. Crusher gradients symmetrically around the 2nd RF pulse and at the end of T_R ensure a consistent signal independent of the background B_0 field.

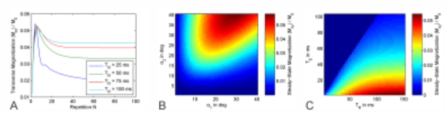

Figure 2: Dependence of the magnetization of sequence parameters. If not noted differently, simulations were performed using α_1/α_2 = 15°/30°, T_E = 12 ms, T_R = 100 ms and T_1/T_2 = 900/50 ms. A) Transverse magnetization reaching steady-state for different values of T_R. With longer T_R the steady-state is reached sooner and the steady-state magnetization is larger. B) Steady-state magnetization subject to the flip angles α_1 and α_2. C) Steady-state magnetization subject to echo time T_E and repetition time T_R. Longer T_R and shorter T_E lead to higher steady-state magnetization.

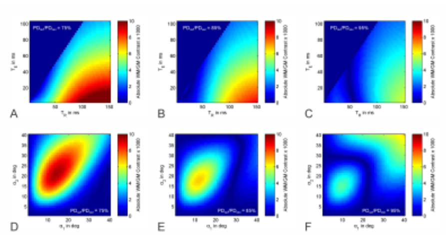

Figure 3: Absolute contrast between GM and WM. A-C) Contrast dependency on the sequence timings for different relative PD of the two tissue types at α_1/α_2 = 10°/20°. Contrast increases with shorter T_E and longer T_R, however, there is a band of low contrast at intermediate values of T_R that moves to shorter T_R with lower relative PD. D-F) Contrast shows an area of maximum values around α_1/α_2 = 10°/15° with larger peak values for smaller values of relative PD (T_E/T_R = 12/100 ms).

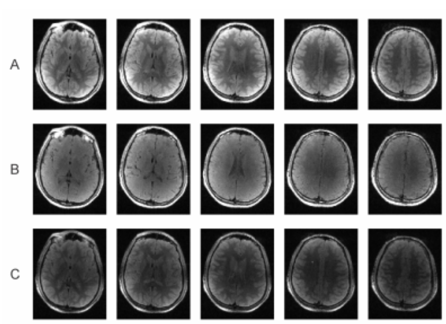

Figure 4: In vivo LACE images of a healthy human brain (FoV 170x210x128 mm^3 at 2x2x4 mm resolution). Shown are 5 central slices acquired at A) α_1/α_2 = 10°/20° and T_E/T_R = 12/100 ms, B) α_1/α_2 = 10°/40° and T_E/T_R = 12/100 ms, and C) α_1/α_2 = 10°/20° and T_E/T_R = 12/50 ms. In line with the simulation results at a relative PD of about 85% (Fig. 3), A) shows the highest GM/WM contrast, while B) shows largely no contrast and C) shows low contrast and reduced signal strength.

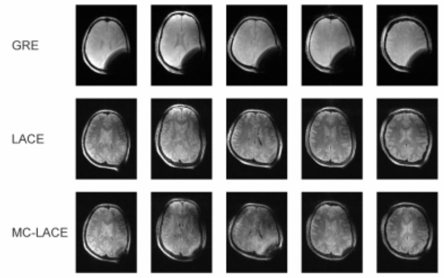

Figure 5: LACE brain images of 5 human subjects in the presence of B_0 inhomogeneity induced by a paper clip (FoV 256x192x128 mm^3 at 2x2x4 mm resolution). 3D gradient echo images show large areas of signal dropout near the paper clip (top row). LACE images recover signal in this area, but exhibit image deformations due to the extreme inhomogeneity (middle row). Multi-Coil generated LACE images show largely the same image quality than the ones acquired with standard gradient coils (lower row). No image distortion correction was applied.

Proc. Intl. Soc. Mag. Reson. Med. 29 (2021)
0004