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Original contributions

Quantitative SENSE-MRSI of the human brain ☆

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Abstract

Purpose: To develop a method for estimating metabolite concentrations using phased-array coils and sensitivity-encoded (SENSE) magnetic resonance spectroscopic images (MRSI) of the human brain.
Materials and Methods: The method is based on the phantom replacement technique and uses receive coil sensitivity maps and body-coil loading factors to account for receive B_1 inhomogeneity and variable coil loading, respectively. Corrections for cerebrospinal fluid content from the MRSI voxel were also applied, and the total protocol scan time was less than 15 min. The method was applied to 10 normal human volunteers using a multislice 2D-MRSI sequence at 3 T, and seven different brain regions were quantified.
Results: *N*-Acetyl aspartate (NAA) concentrations varied from 9.7 to 14.7 mM, creatine (Cr) varied from 6.6 to 10.6 mM and choline (Cho) varied from 1.6 to 3.0 mM, in good general agreement with prior literature values.
Conclusions: Quantitative SENSE-MRSI of the human brain is routinely possible using an adapted phantom-replacement technique. The method may also be applied to other MRSI techniques, including conventional phase encoding, with phased-array receiver coils, provided that coil sensitivity profiles can be measured.

Keywords: Brain; Magnetic resonance; Spectroscopic imaging; SENSE; Quantitation

1. Introduction

Proton magnetic resonance spectroscopic imaging (MRSI) of the brain has been used widely to characterize regional variations of spectra in the healthy brain during development and in adulthood, as well as for examination of genetic, neurodegenerative, psychiatric, inflammatory and neoplastic pathological conditions [1,2]. MRSI allows for a more efficient acquisition of spectra from multiple regions as compared to single-voxel magnetic resonance spectroscopy and has the ability to depict spectral information as images that can be directly compared to magnetic resonance imaging (MRI). Spatial localization in MRSI is primarily dependent on phase encoding. Generally, phase encoding of any dimension will extend the scan time by a factor equal to the number of phase-encoding steps. Typically in MRSI, at least two dimensions are phase encoded, making data acquisition time consuming, particularly when both high spatial resolution and large volume coverage are required. Therefore, there have been efforts to reduce MRSI scan time using various different approaches (many of which were first developed for fast MRI experiments). Some of the fast MRI techniques that have been adapted for use in MRSI include echo train acquisition [3], echo planar techniques [4], reduced *k*-space acquisition [5,6] and parallel imaging methods [7,8].

Sensitivity encoding (SENSE) is a promising parallel imaging technique for scan time reduction in ^{1}H MRSI of the human brain [7,9]. The technique allows the reduction of *k*-space sampling density, and thus scan time, by using the spatial sensitivity variations of multiple receiver coils for additional spatial encoding. With appropriate reconstruction algorithms, the combination of the multiple receiver coil data allows for the creation of spectroscopic images without the presence of aliasing artifacts that would otherwise result from the undersampling of *k*-space data [7].

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Until recently [10], one limitation has been a lack of quantitation methods for SENSE-MRSI. Existing quantitation methods for MRSI [11] include external reference techniques, either the phantom replacement technique [12] or the use of phantoms placed within the coil during scanning, and internal reference methods, such as the use of the water peak [13]. These methods have been used previously for conventional MRSI data recorded with a single transmit–receive head coil. The phantom replacement technique appears to be one of the better techniques, since it does not require additional patient scan time, and it involves a standard of known concentration, unlike water referencing methods which require assumptions about water content and relaxation times [10,14]. External standards also suffer from problems, including B_0 magnetic susceptibility effects (causing decreased field homogeneity in adjacent brain), sensitivity to B_1 inhomogeneity and the requirement of larger field of views (FOV) than would otherwise be used. While the phantom replacement method avoids some of these problems, it is sensitive to variable coil loading and B_1 inhomogeneity, factors which are not straightforward to estimate for phased-array receiver coils which are required for SENSE encoding. Both transmit and receive B_1 fields are sensitive to coil loading effects which usually vary between the brain and the phantom, and from one subject to another.

This work describes the development of a quantitative multislice SENSE-MRSI protocol at 3.0 T based on the phantom replacement technique, with B_1-field estimation to account for variability of the receiver coil fields, and loading correction performed via the body coil RF transmitter voltage. Data are presented for seven selected brain regions in 10 normal volunteers and include corrections for tissue relaxation times and partial volume with cerebrospinal fluid (CSF). It is demonstrated that reliable metabolite concentrations can be recorded with a rapid SENSE-MRSI protocol that are in concordance with previous values measured using conventional MRS methods.

2. Materials and methods

2.1. Phantom replacement with phased-array coils

The phantom replacement technique relies on a phantom of known concentration to be scanned immediately before or after the in vivo scan [12]. If the MR system is stable over time, it is not necessary to perform the reference scan for every subject. However, in order to be able to compare signal intensities between the reference and in vivo scans, care must be taken to duplicate the scanning parameters for each study. The same data acquisition parameters, spatial resolution and postprocessing procedures must be used [12]. However, some differences between the reference and in vivo scans usually cannot be avoided; most likely the coil loading will be different, and the receiver gain may be different, and the phantom relaxation times will be different from those in vivo. These factors all need to be corrected for in the quantitation procedure.

A previous implementation of quantitative MRSI using the phantom replacement technique used a volumetric, transmit–receive birdcage coil which was assumed to be homogeneous over the brain volume covered [12]. By the principle of reciprocity [15], the coil loading correction factor (to be applied to the received signal) was calculated from the power required to perform a 90° pulse. However, for phased-array receiver coils, the B_1 field is spatially inhomogeneous, different coil elements may load differently and a separate transmitter coil is normally used, such that the receive coil load factors cannot be simply deduced from the transmitter voltage. Fortunately, both the B_1 inhomogeneity and the variable receiver coil loading can be measured (as described below) relative to the body coil, for which a load factor can be determined, since it is a transmit–receive coil. This procedure has previously been applied for single-voxel spectroscopy of the brain with phased-array coils [16,17], but not previously for quantitative MRSI as far as we are aware.

Formally, for a system with multiple (n=1 to N) receiver coils and a separate transmitter coil (often the scanner 'body' coil), the detected signal from a metabolite of concentration [M] in the nth channel can be expressed as

$$S_n(x,y) = g_n k_n B_{1n}(x,y)[\mathrm{M}] / F_n \quad (1)$$

where k_n is a proportionality constant dependent on numerous factors, including the pulse sequence used, metabolite relaxation times and other instrumental settings, and g_n is the receiver gain of the nth channel [15]. $B_{1n}(x,y)$ is the magnetic field that would be produced at point (x,y) by a unit current flowing through coil n in the absence of any coil loading, and F_n is the 'load factor' which describes how the signal is modified in the presence of a sample. Note that in this article, the quantity F is operationally defined to include both the true loading effect of the sample on the coil (i.e., the decrease in coil quality factor Q), as well as any change in signal which is caused by a variation in RF coil impedance (which could potentially be corrected by retuning and matching the coil from sample to sample, but which is usually not performed on clinical MRI scanners).

If the transmitter coil were also to be used as a receiver, then the signal from this coil (S_t) is given by the analogous expression $S_t(x,y)=g_t k_t B_{1t}(x,y)[\mathrm{M}]/F_t$, where F_t is linearly proportional to the transmitter voltage required to generate a slice-selective 90° pulse.

The phantom replacement quantitation technique compares the signal from a standard sample of known concentration designated [P], with that of the metabolite to be determined [M], where separate in vivo (i) and phantom (p) experiments are performed. As has been shown