RECENT ADVANCES IN IMPEDANCE AND CURRENT MR IMAGING

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ABSTRACT

Magnetic resonance imaging (MRI) techniques have become important tools in medicine, particularly for clinical diagnosis. Conventional MRI, however, does not reveal information about the electrical properties of the living body. New methods for imaging impedance and electrical currents in the body based on new principles of MRI including the large flip angle method, additional time-varying magnetic field method, diffusion tensor method, direct neuronal current MR imaging, and the resonant frequency shift technique are introduced and reviewed. These new imaging techniques will be useful for the investigation of the higher brain functions of the human brain in the future.

INTRODUCTION

Magnetic resonance imaging (MRI) techniques have become important tools in medicine, particularly for clinical diagnosis. Conventional MRI, however, does not provide information about the electrical properties of the living body. This paper reviews and proposes new methods for the imaging of impedance and electrical currents in the body based on new principles of MRI.

IMPEDANCE MR IMAGING

Since electrical properties are important characteristics of living organisms, techniques for impedance tomography to visualize impedance distribution have been developed with great interest. The previously proposed techniques require the attachment of electrodes to the surface of the human body.

When conductive tissues are subjected to time-varying magnetic fields in MRI, eddy currents are induced. A new method for impedance tomography based on MRI techniques was achieved by applying the effects of the eddy currents on spin precession. In addition, another type of impedance MR imaging based on diffusion tensor MRI was achieved by measuring the diffusion rate of ions and water molecules in living tissues that varies with the tissue conductivity. We introduce three types of impedance imaging techniques based on MRI.

IMPEDANCE MRI BASED ON THE LARGE FLIP ANGLE METHOD

When conductive tissue is exposed to RF magnetic fields, eddy currents are induced in the conductive tissues, which results in the reduction of the net RF fields into the tissue. By the shielding effects of eddy currents on spin precession, the flip angles, (i.e., mutation angles of the macroscopic magnetization of excited spins from the axis of the main static field Bo) are reduced in varied degrees, depending on the electrical characteristics of the tissue [1].

When a precise 180°, 360°, or 540° excitation pulse is applied to conductive tissue, the tissue does not yield a signal, due to absence of the transversal components of magnetization. Meanwhile, resistive tissues yield a signal because it is less electrically shielded than conductive tissue and simultaneously undergoes a different flip angle. Also, resistive tissue maintains a transverse component, with magnitude determined by the sine-wave function of the flip angle. The difference in signal, therefore, reflects the conductivity of tissue. By applying very large flip angles; conductivity-enhanced MR images can be obtained, yet, only at the given Larmor frequency, and in the direction perpendicular to the applied rf field.

IMPEDANCE MRI BASED ON THE ADDITIONAL TIME-VARYING MAGNETIC FIELD METHOD

Another proposed method is to apply an additional time-varying magnetic field parallel to the main static field B_0. A third coil, B_c coil, produces the additional magnetic field. The method is used to obtain conductivity-enhanced MR images at an arbitrary frequency. When the perturbing field or B_c field is applied, slice positioning of the image is affected, and slice selection fluctuates. Spatial information in the read-out and phase-encoded directions is also affected. Due to the shielding effects, the B_c field does not affect the conducting tissue that much. Since the frequency of B_c field is independent of the given Larmor frequency, conductivity-enhanced images can be obtained at any frequency, except in the direction perpendicular to the B_c field [1-2].

IMPEDANCE MRI BASED ON THE DIFFUSION TENSOR METHOD

When an ion with a charge q moves in a solution by an applied field E, the balance of the electrostatic force and the viscous drag is expressed as $qE=6\pi r_i\eta v$ (1), where r_i is the Stokes radius of the ion, η is the viscosity of the solution, and v is the drift velocity of the ion. The current density j in the solution of the ion density N is $j=qNv$ (2). The self-diffusion coefficient of water is $D=kT/6\pi r_w\eta$ (3), where k is the Boltzmann constant, T is the temperature, and r_w is the Stokes radius of water molecules. Thus, the relationship between the conductivity σ and the self diffusion coefficient D is obtained from (1)(2)(3) as $\sigma=j/E=(r_wq^2N/r_ikT)D$ (4).

Diffusion MR images of a rat brain were acquired in a transversal slice. Motion probing gradients (MPGs) were applied with a b-factor of $1.4\text{x}10^9$[s/m^2] in the read-out and phase-encoding directions. The self diffusion coefficient of water was calculated in each pixel as $D=(1/b)\ln(S_0/S)$, where b is the b-factor, S_0 and S was the signal intensity of the images with and without a MPG. Conductivity maps were generated by equation (4) with the following variables, $r_w/r_i=0.76$, $q=1.6\text{x}10^{-19}$[C], $N=2.0\text{x}10^{25}$[/m^3], $kT=4.1\text{x}10^{-21}$[J]. Hence, impedance imaging with anisotropic characteristics in different directions was obtained based on diffusion tensor MRI techniques [3].

RESULTS AND DISCUSSION OF IMPEDANCE MRI

Experiments were carried out using mice and rats in a 7.05 T, 300 MHz, 13.3 cm diameter MRI system. The MR image at a 180° flip angle showed a small signal from the brain and muscle tissues because there were almost no transversal components of magnetization. On the other hand, in the same 180° image, the resistive fatty tissues, which were transparent to the RF field, yielded a specific signal. By applying 180° pulses to the conducting CSF and muscle tissues, resistive fatty tissues simultaneously received excitation of flip angles larger than 180° and produced an image signal.

The method of additional time-varying magnetic fields of a 1 KHz sinusoidal wave produced low quality impedance MR images as compared with the images of the large flip angle method.

Both the large flip angle method and the additional time-varying Bc field method are sensitive to RF inhomogeneity. In addition, both the Larmor frequency B1 field and the low frequency Bc field are easily transmitted, absorbed, and reflected by biological tissue boundaries in varying degrees. The variability depends on the geometry of the subject, tissue properties, frequency, and the direction of the B1 or Bc field.

The obtained T_1-weighted image of the rat brain showed horizontal (phase-encoding) and vertical (read-out) conductivity maps. The skull does not affect this method of conductivity mapping. In addition, the anisotropy of the conductivity can be investigated by changing the direction of the MPG. Current distributions in the electromagnetic stimulation of the brain can be calculated from the conductivity maps as well. As an example, we calculated the current distributions in the rat brain when a current was applied by a pair of electrodes placed on the upper edge and the lower edge of the model. This new method of conductivity imaging based on diffusion MRI provides calculation models of biological tissues with inhomogeneous and anisotropic conductance at high spatial resolutions.

DIRECT NEURONAL CURRENT MRI

Although functional MRI (fMRI) has become a powerful technique for studying human brain function, it cannot directly detect the electrical activities of neurons in the brain. This is because fMRI uses the information of the local spin-spin interaction T2* based on the blood oxygenation level dependent (BOLD) effect. In other words, T2* can detect the change of the magnetic susceptibility of oxy-deoxy hemoglobin in blood vessels. Therefore, fMRI can detect brain function indirectly via the information of cerebral blood flow [4].

Direct neuronal current MRI can directly image the spatial distribution of neural currents in the brain. The basic principle is to erase the effects of BOLD by subtracting MRI signals with different polarities of gradient magnetic fields [5].

We obtained the first ever direct neuronal current image of a human brain during finger-tapping exercises. Although the sources appeared in the somatosensory and parietal areas rather than in the expected motor cortex area, this new imaging technique will be useful for the investigation of the higher brain functions of the human brain in the future.

CURRENT MR IMAGING BASED ON THE RESONANT FREQUENCY SHIFT TECHNIQUE

Another new method to detect electrical currents is based on the resonant frequency shift technique of MRI. The resonant frequency shift method has been used for MR spectroscopy; however, it has never been applied to detect electrical currents. We introduced a loop current and a straight-line current as models of electrical sources in the brain, nervous and muscular systems. We detected changes of magnetic fields generated by the loop current and the straight line current by obtaining resonant frequency shift images [6].

When an electrical current was applied to a 3-turn loop cable and a straight-line cable that were placed in spherical phantoms filled with water during image acquisition, the magnetic fields generated around the cable were added to the main magnetic field. The main magnetic field strength increases or decreases around the electric currents, which causes a shift in the resonant frequency. A resonant frequency shift sequence based on the spin-echo (SE-CHESS, Spin Echo Chemical Shift Selective) was used. In the resonant frequency shift sequence, a 90-degree pulse was applied with a narrow bandwidth (160Hz) so that the spins with a shifted resonant frequency were not excited. A slice selective gradient was added with only a 180-degree pulse instead of a 90-degree pulse. Gaussian RF pulses were used so that the decrease of the signal intensity corresponded with the amplitude of the magnetic field generated around the current. The parallel component of the generated magnetic field causes a detectable resonant frequency shift. The spins are not sufficiently excited in the region where the resonant frequency changes in the phantom, resulting in a decrease of signal intensity. By detecting the decrease in the MR signal, we can observe the change of the magnetic field generated by the current.

The obtained images of the loop current phantom revealed a decrease of the signal intensity inside the loop. The signal intensity decreased when the magnetic fields generated by the loop current were parallel to the main static field. The obtained images of the straight current phantom showed that the signal intensity decreased at the region where the straight current generated a magnetic field and where the magnetic field was parallel to the main magnetic field direction.

A decrease of the signal intensity inside the loop was observed. The signal intensity decreased when the magnetic fields generated by the loop current were parallel to the main static field. Magnetic field maps are obtained by evaluating the decrease of the signal intensity. The magnetic field map can identify the amplitude and position of the current.

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