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Magnetic resonance imaging of mechanical deformations

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ABSTRACT

A method for magnetic resonance imaging of mechanical deformations is presented. The method utilizes an MRI compatible device for inducing elastic deformations of a sample and a modified spin-echo imaging sequence with two position-encoding gradients added to the sequence symmetrically to the RF refocusing pulse. At the end of the first position-encoding gradient pulse, a sample deformation was induced by the deformational device, which applied a force to a plastic rod embedded in a gelatin cylindrical sample. The sample had to withstand repeated elastic deformations. Sample displacements up to 400 μm were encoded in the image signal phase by the use of position-encoding gradients. Images of different displacement components were acquired first by the use of position-encoding gradients in different directions and then processed by the 2D phase unwrap algorithm. Finally, images of normal and shear strain distribution were calculated from the displacement images. The obtained displacement and strain images enabled clear visualization of deformations and their extent in the sample with the displacement detection threshold in the range 0.3–0.6 μm, depending on the image echo time. The results of displacements were verified also by a DANTE tagging method and by an optical method. The presented method enables studying of various types of deformations in different soft materials as well as dynamic response of deformations to different stress functions (static, oscillatory, pulsed…).

1. Introduction

Magnetic resonance imaging is a powerful tool for studying various phenomena also due to its ability of providing not only signal magnitude of each image element but also signal phase. In conventional MRI signal phase is often irrelevant, however, there are several methods where the phase plays a crucial role. These include phase contrast angiography (PCA) [1,2], MR thermometry [3,4], MR elastography [5], current density imaging (CDI) [6,7]… In all these methods phase carries information on the measured quantity; in PCA phase is proportional to fluid velocity, in MR thermometry phase is proportional to temperature, in MR elastography phase is proportional to an amplitude of the propagated sound wave, in CDI phase is proportional to a magnetic field change induced by electric currents… The present work is focused to MR imaging of mechanical deformation, which is also another method of phase detection based MRI.

In different materials deformations are measured differently. For example in solids deformations can be measured optically by Moiré interferometry [8], in photoelastic materials deformations can be accessed by the change of the birefringence [9]. Deformations in solids can be imaged also in nanoscale resolution by infrared near-field microscopy if the materials are infrared-active [10]. It is also possible to image large-scale deformations; for example deformations of the Earth's surface, can be imaged by the Synthetic Aperture Radar (SAR), a satellite-transmitted and received radar system [11]. In medical imaging deformations of biological tissues are often imaged by ultrasound. An important application of that is myocardial deformation imaging [12] which is a novel echocardiographic method for assessment of global and regional myocardial function. By MRI, deformations of biological tissues can be assessed by the tagging method, which is a popular method in cardiac MRI for assessing ventricular function. The method was first introduced as a SPAtial Modulation of Magnetization (SPAMM) method [13] and then improved with inclusion of Delay Alternating with Nutations for Tailored Excitation (DANTE) RF pulses that together with an appropriate combination of magnetic field gradients result in a grid pattern of excited spins [14]. Later improvements in the field include Complementary SPAMM (CSPAMM) [15], HARmonic Phase (HARP) [16] and Strain ENCoding (SENC) [17] methods. By the tagging methods the heart motion can be assessed from the grid deformation formed in time between the excitation and image acquisition. MRI enables displacement field measurement also by texture correlation, a technique which relies on unique image patterns within a pair of digital images to track displacement [18]. Another way of resolving myocardial kinematics is by using a phase contrast approach. The approach is used in the Strain Phase MRI (SP-MRI) method [19] in which sample displacements are encoded by a readout gradient of a

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conventional spin-echo sequence. Larger encoding times are enabled by the Displacement ENcoding with Stimulated Echoes (DENSE) method [20,21], which is used primarily in cardiac functional MRI. In MR elastography [5] acoustic sound is emitted to a sample synchronously with oscillating gradients that have the same frequency as the sound which results in a phase shift proportional to the sound amplitude in the sample. As the sample is dynamically deformed during sound propagation in it, MR elastography can also be considered as a way deformation imaging.

Motivation for this study was strain imaging of elastic materials with a precise displacement detection. Sensitivity of one micrometer at a spatial resolution of the order of 100 μm is desired. As most of the abovementioned methods are optimized for cardiac imaging where displacements are two orders of magnitude larger, an improved detection method is needed. For that we propose a spin-echo imaging sequence to which two short and strong position-encoding gradient pulses are added, i.e., a pulsed gradient spin-echo (PGSE) sequence. The first gradient pulse encodes sample's initial position, while the second gradient pulse encodes sample's final position. In between the gradient pulses the sample is deformed by applying an external force to it in a pulse of variable duration. This approach enables imaging of deformation progression as a function of the applied force pulse duration as well as observation of transition phenomena, such as sample oscillations or shock wave propagation that follow the pulse. First, theoretical background of the method along with a pulse sequence for deformation imaging is presented. This is then followed by a demonstration of the method on a gelatin cylindrical sample that was repeatedly deformed by force pulses generated by a special MRI compatible deformational device. Finally, the measured data were analyzed to obtain sample displacement and strain maps. For the study a special MRI compatible deformational device was designed that enables full control over the applied force and its duration. The results of displacements were also verified by a DANTE tagging method and by an optical method based on laser beam reflection.

2. Materials and methods

Imaging of mechanical deformations was tested on a gelatin sample that was deformed by an MRI compatible device shown in Fig. 1. The device consisted of a flat round coil with $N = 60$ turns of copper wire and with a diameter of $2r = 34$ mm. The coil was attached to a 94 mm long shaft to the center of which was glued an $l = 23$ mm long lever arm. The arm was joint with a 90 mm long moving rod embedded in the gelatin sample. During sample preparation, the rod was immersed to the depth of 14 mm in a 25 mm thick gelatin layer during its hardening in a container (glass beaker) with inner diameter of 24 mm. Gelatin ("Želatina", Dr. Oetker Kft, Janossomorja, Hungary) was prepared according to the manufacturer's instructions; 5 g of gelatin powder was admixed to 50 ml of water and then cooked. NMR relaxation times of the sample were shortened to $T_1 = 266$ ms and $T_2 = 155$ ms by adding the mixture 0.1 g of copper sulfate. The contact between the rod and gelatin was improved by sanding the embedded part of the rod. When gelatin was hardened the sample was inserted in a 27 mm micro-imaging probe (Bruker, Ettlingen, Germany). The rod was then connected with the lever arm of the deformational device and the device was attached to the probe holder and inserted into a 100 MHz ($B_0 = 2.35$ T) horizontal bore superconducting magnet (Oxford Instruments, Abingdon, UK).

Deformations of the sample were induced by applying current of $I_0 = 14$ mA to the coil. As the coil was in magnetic field of the MRI magnet and the coil axis was normal to the magnetic field direction, the current induced a torque to the coil equal to $\tau = N\pi r^2 I_0 B_0 = 0.0018$ Nm. The torque initially resulted in an acceleration of the moving rod equal to $a_0 = \tau l/I = 23$ m/s^2 (calculated from geometrical parameters of the deformational device); here $I = 1.74 \cdot 10^{-6}$ kg/m^2 is a net moment of inertia of all moving parts of the deformational device. Finally, when the rod came to a rest, the torque resulted in a force $F = \tau/l = 0.08$ N of the moving rod to the gelatin sample that deformed the sample. Current I_0 was generated by an amplifier that was controlled by TTL pulses (line I of pulse sequences in Fig. 2) of a spectrometer (Tecmag Apollo, Houston, TX, USA). As the coil had low resistance and inductance ($R = 2.6\ \Omega$, $L = 150$ μH) change from zero current to constant current I_0 was reached only after a few microseconds. During the experiment temperature of the gelatin sample was monitored by a copper-constantan thermocouple thermometer inside the micro-imaging probe.

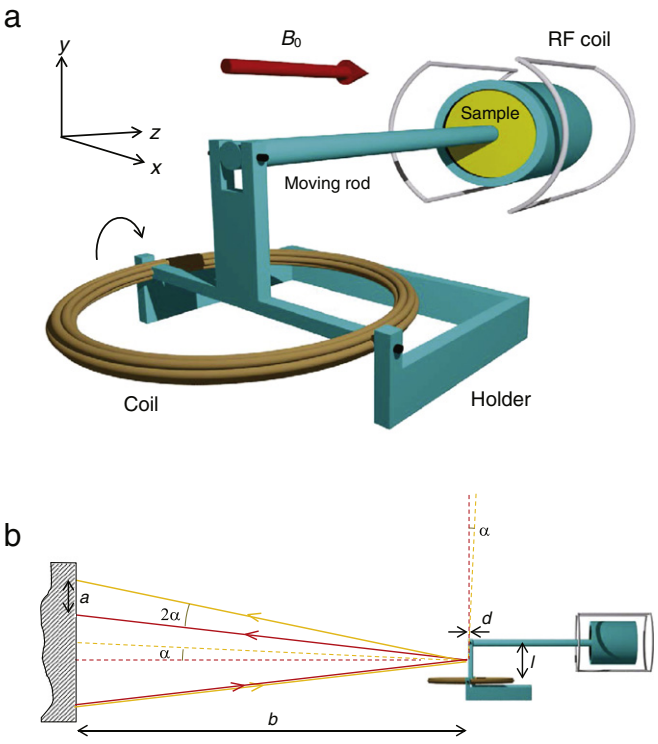

Fig. 1. A scheme of the deformational device (a) and a setup for displacement measurements by a laser beam reflection (b). The device was placed in magnetic field of the NMR magnet. A current pulse applied to the coil therefore resulted in a torque to the coil that was converted via the lever arm to a force on the moving rod (embedded in the gelatin sample) that deformed the sample.

A PGSE sequence, shown in Fig. 2a, was used to image sample deformations. To a spin-echo sequence two position-encoding short and strong gradient pulses positioned symmetrically to the refocusing RF pulse were added. The pulses had a duration $\delta = 2$ ms, amplitude $G = 200$ mT/m and were apart for different deformation times $\Delta = 6$, 10, 20, and 30 ms. The function of the pulses was to encode in a signal phase the initial $\vec{r}_1$ and the final $\vec{r}_2$ position of a sample volume element. The two gradient pulses had an opposite effect to the signal phase shift due to the RF refocusing pulse in between them so that a phase shift of the signal from the element was equal to

$$\Delta\varphi = \gamma\delta\,\vec{G}\cdot\vec{r}_2 - \gamma\delta\,\vec{G}\cdot\vec{r}_1 = \gamma\delta\,\vec{G}\cdot\vec{u}\,. \quad (1)$$

Here $\vec{u} = \vec{r}_2 - \vec{r}_1$ is the element's displacement that is formed in the time interval Δ by the activation of the deformational device at the end of the first position-encoding gradient pulse (line I in