Available online at www.sciencedirect.com
SCIENCE DIRECT®
Journal of Magnetic Resonance 169 (2004) 308–312
JMR Journal of Magnetic Resonance
www.elsevier.com/locate/jmr

Surface normal imaging with a hand-held NMR device

M.C.A. Brown,[a,*] D.A. Verganelakis,[b] M.J.D. Mallett,[c] J. Mitchell,[d] and P. Blümler[e]

[a] School of Physical Sciences, University of Kent, Canterbury, CT2 7NR, UK
[b] Department of Chemical Engineering, University of Cambridge, Cambridge CB2 3RA, UK
[c] Oxford Magnet Technologies Ltd., Witney, Oxfordshire OX29 4BP, UK
[d] School of Electronics and Physical Sciences, University of Surrey, Guildford GU2 7XH, UK
[e] Max-Planck Institute for Polymer Research, D-55128 Mainz, Germany

Received 7 November 2003; revised 9 April 2004
Available online 11 June 2004

Abstract

Recently the capabilities of single sided nuclear magnetic resonance (NMR) devices have been extended towards three-dimensional imaging. This paper details the use of a magnetic field sweep coil to obtain spatial resolution in the plane normal to the surface of a hand-held NMR device—the NMR-Mobile Universal Surface Explorer (MOUSE). One-dimensional depth profiles can be recorded by varying the current in the sweep coils. Preliminary results from multi-layer rubber and glass sample phantoms demonstrate a sample penetration depth of 7 mm. Two-dimensional images were acquired via the inclusion of phase encoding coils. Non-destructive cross-sectional images of small rubber phantoms were successfully recorded.

Keywords: NMR-MOUSE; Single sided NMR; Portable MRI; Imaging; Magnetic field sweep

1. Introduction

Portable single-sided nuclear magnetic resonance (NMR) probes have several advantages over conventional NMR systems. They offer open access to the sample space, are low in cost, and robust. The main application of these devices has been the non-destructive study of self-diffusion and relaxation phenomena in volumes close to the surface of the sample [1]. Non-destructive imaging of samples using single-sided NMR devices offers many more biomedical and industrial applications. The device discussed in this paper has been based on a hand-held probe: the NMR-Mobile Universal Surface Explorer (MOUSE)[1] [2], with adequate improvements [3] for one- and two-dimensional imaging. This modified device has been named the magnetic resonance imaging (MRI) MOUSE by the authors. The nature of the probe prohibits the application of conventional linear field gradients perpendicular to its surface. The B_0 magnetic field of the MOUSE already has a gradient in the order of 10–100 T/m in this direction and so any applied radio frequency (RF) excitation pulses will automatically be spatially selective. Therefore the observed NMR signal originates from a thin "slice" approximately parallel to the surface of the, and whose displacement is controlled via the proton resonant frequency [4]. This sample "slice" is known as the "resonant volume" of the MOUSE. Early images from the axis normal to the MOUSE surface were obtained by manually tuning the resonant circuit to a succession of frequencies using a selection of fixed capacitors [5]. This had the effect of moving the resonant volume through a series of discrete elevations above the surface of the sensor. One drawback of this method was the time consuming manual retuning of the resonant circuit that was not necessarily reproducible. The results from such experiments required careful calibration to allow signals from different sample depths to be directly compared.

Now a continuous, computer controlled method of depth penetration has been achieved by employing an additional time-dependent magnetic field produced by a

* Corresponding author. Fax: +44-1227-827558.
E-mail address: mcab2@kent.ac.uk (M.C.A. Brown).
[1] NMR-MOUSE is a registered trademark of RWTH-Aachen, Germany.

1090-7807/$ - see front matter © 2003 Published by Elsevier Inc.
doi:10.1016/j.jmr.2004.04.013

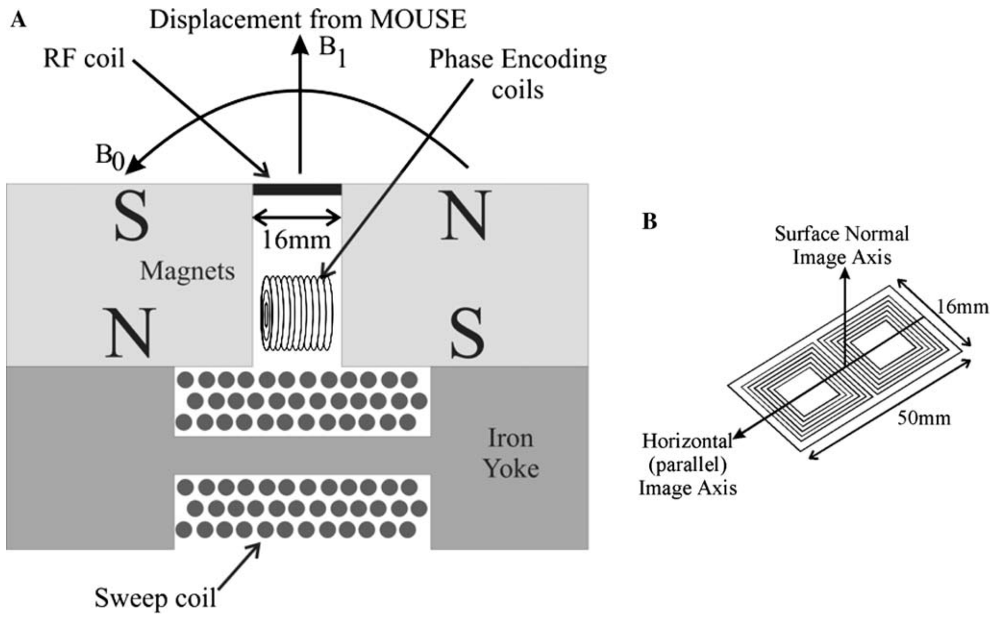

Fig. 1. (A) Schematic cross-section of a 2-D MRI-MOUSE, including a sweep coil to provide vertical spatial resolution, and phase encoding coils to provide horizontal spatial resolution. Both imaging directions are normal to the B_0 field. This provides a two-dimensional imaging plane in the centre of the MOUSE. (B) Axes of image orientation relative to RF surface coil, providing a two-dimensional image in a plane normal to the surface of the MOUSE.

magnetic field sweep coil [6]. This coil was integral to the design of the MRI-MOUSE. By varying the magnetic field rather than the resonant frequency, a uniform signal is obtained at greater depths. By combining this technique with regular phase encoding coils to provide resolution parallel to the surface of the MOUSE, two-dimensional images have been obtained including T_2 relaxation time weighted image contrast. Full details on the design, construction, and implementation of the MRI-MOUSE can be found in [7].

2. The MRI-MOUSE

The MRI-MOUSE (Fig. 1) construction is based on the palm-sized MOUSE designed by Blümler and Blümich [2]. The total dimensions of the MOUSE are $96 \times 50 \times 50$ mm (length $\times$ width $\times$ height). The basic MOUSE consists of two permanent magnets of dimensions $40 \times 50 \times 25$ mm. They are made of highly magnetised iron neodymium boron (FeNdB), providing a B_0 field of 0.45 T at the surface of the MOUSE. The magnets are mounted in antiparallel positions leaving a 16 mm horizontal gap between them, and are joined underneath by an iron yoke that channels the magnetic flux. A "figure-8" RF surface coil [3] was mounted in the gap between the magnets. The B_1 field is projected normal to the surface of the MOUSE and is perpendicular to the B_0 field.

A magnetic field sweep coil was added to the basic MOUSE design to obtain spatial information above the surface of the MOUSE. The coil consisted of approximately 350 turns of 0.5 mm diameter lacquered copper wire, wound round the iron yoke below the gap between the two magnets; see Fig. 1A. The coil was located on the yoke where the thermal mass of the iron was sufficient to prevent the magnets being damaged by heat evolution in the coil during the experimental time. When supplied with a current the sweep coil generates a magnetic field that augments or reduces the existing static B_0 field. Here we use the convention that a positive current produces a magnetic field that increases B_0. A field increase will cause the resonant volume to move away from the surface of the MOUSE. Since the proton resonant frequency ω_0 is fixed, the resonance condition, $\omega_0 = \gamma B_0$ is fulfilled at different positions. Conversely a negative current will move the resonant volume toward the MOUSE.

Two phase encoding coils, connected in series, were mounted horizontally (aligned along the B_0 field) just below the RF coils (see Fig. 1A) and a copper shield was added to eliminate RF noise induced in the resonant circuit. These phase encoding coils were also made from lacquered 0.5 mm diameter copper wire, counter-wound into multi-layer solenoids, each comprising of 100 turns in total. The phase encoding coils provided a magnetic field gradient that varied across the long axis of the RF coils; see horizontal (parallel) image axis in Fig. 1B.

The imaging experiments performed using the MRI-MOUSE were controlled by a Maran Ultra spectrometer manufactured by Resonance Instruments.[2] The pulse sequence employed was a θ_y–$(\tau$–θ_x–τ–echo$)_n$ sequence

[2] Resonance Instruments, Whitney, Oxfordshire, UK.

[8], similar to a standard CPMG pulse sequence [9,10], where was the optimised spin flip angle that provided the largest initial echo amplitude. The flip angles from the applied pulses were not precisely defined due to the spatial inhomogeneity in both B_0 and B_1. The first excitation pulse flips the spins onto the x-axis. The second excitation pulse is applied along the x-axis after a time τ to approximately rotate the spins into the x–z plane so they refocus along the x-axis, forming a detectable echo at time 2τ. The second excitation pulse is repeated at intervals of 2τ to form a train of echoes. The pulse sequence used a CYCLOPS phase cycle, and all the alternate echoes were co-added to improve signal-to-noise. The sweep gradient is applied constantly, but the phase gradient is only active between the initial two excitation pulses. During the collection of 1000 echo trains, at a repeat delay of 1 s, the temperature of the sweep coil increased by 20 °C.

A Techron[3] 8300 series class A/B audio amplifier, regulated by a Resonance Instruments gradient control card, drove the sweep coil. The current through the sweep coil could be incremented from −24 A to +24 A, using up to 60 steps, depending on the required resolution. A current of +24 A corresponded to a field strength increase of 160 mT in B_0 at the surface of the RF coil. However, it was discovered that the permanent magnetic field of the MOUSE had already caused magnetic saturation in the iron yoke. Therefore, additional magnetic flux from the sweep coil could not be used to augment the B_0 field. Instead, only negative sweep currents could be applied to reduce the B_0 field. Accordingly the proton resonant frequency was set at 12 MHz which corresponded to a volume 7 mm above the surface of the MOUSE, determined from the known B_0 field profile. When a sweep current of −20 A was applied, the resonant volume moved down to be adjacent to the surface of the RF coil. There was an inherent loss in sensitivity caused by the use of a lower resonant frequency. A larger yoke could have been fitted to allow positive sweep currents and a higher resonant frequency to be used. However, the inclusion of more iron would greatly increase the weight of the device and reduce its portability. Another Techron amplifier was used to provide the phase encoding gradients. For each sweep current step, a standard phase encoding sequence [11] was run to capture a horizontal one-dimensional profile. The maximum safe gradient currents were applied at all times to provide the best horizontal resolution.

Calibration experiments were conducted on a series of phantoms consisting of Otoform[4] (proton-rich rubber) and glass layers. All the experiments were conducted at room temperature (~20 °C), inside a mobile Faraday cage to reduce external environmental noise. The phantoms were designed to determine the resolution and penetration depth of the sweep system. The profile depths were found to have a non-linear dependence on the applied sweep current. This non-linear relationship was used to re-scale all the subsequent measurements. The system was found to have a resolution limit in the order of 2 mm. This relatively poor resolution is partly due to the curved shape of the resonant volume [4]. Methods for improving this resolution have been considered elsewhere [12]. Measurements on a uniform block of Otoform indicated that only sweep currents between −10 and −20 A influenced the position of the resonant volume. The number of sweep current steps used in all subsequent measurements was reduced to this operating range.

3. Experimental results

Three-dimensional Otoform phantoms were constructed to test the two-dimensional imaging capabilities of the MRI-MOUSE. The raw cross-sectional image from one such phantom, a π shape, can be seen in Fig. 2A. A binary image has also been shown that was created by applying a non-linear normalisation function to the raw data. This function was determined from an image of a solid rubber phantom. The true sample cross-section has been illustrated, and the imaged area outlined in Fig. 2B. The image plane was normal to the surface of the MOUSE and along the long axis of the RF coil and has been sketched in Fig. 2C for clarity. The image is a reasonable representation of the actual phantom, providing a non-destructive cross-section. The horizontal dimensions of the image are limited by the shape of the B_1 field and hence the size of the RF coil. A third dimension could, in principle, be imaged by the inclusion of a second set of phase encoding coils, but the field of view would be extremely narrow. Rather, three-dimensional images could be generated simply by mechanically moving the entire MOUSE across the surface of the sample. T_2 weighted images were also produced, demonstrating the ability of the MRI-MOUSE to distinguish materials with different relaxation times. To demonstrate this, a glass and two rubber (Otoform, $T_2 = 80$ ms and Blu-Tack, $T_2 = 40$ ms) π phantom was produced. The three materials were distinguishable in the image due to their different relaxation times; see Fig. 3.

4. Conclusions

In this paper it has been successfully demonstrated that two-dimensional images can be obtained from a

[3] Techron Division of Crown International, Elkhart, IN, 46517-4095, USA.

[4] Otoform-K2; Condensation-vulcanising silicone impression material, supplied by Dreve Otoplastik GmbH, 59423-Unna, Germany.

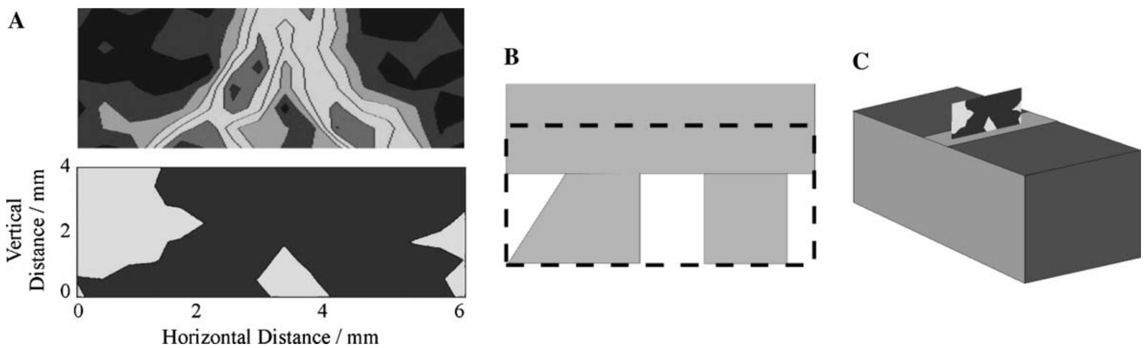

Fig. 2. Two-dimensional, surface normal image recorded using the MRI-MOUSE. (A) The non-destructive, cross-sectional image obtained from the Otoform phantom (top), shown normalised and with a threshold (bottom) for clarity. The true shape is illustrated in (B). The dotted lines in (B) outline the imaged portion of the 'pi' phantom. The sketch in (C) illustrates the orientation of the image on the surface of the MRI-MOUSE, normal to, and on the long axis, of the RF coil; see Fig. 1B.

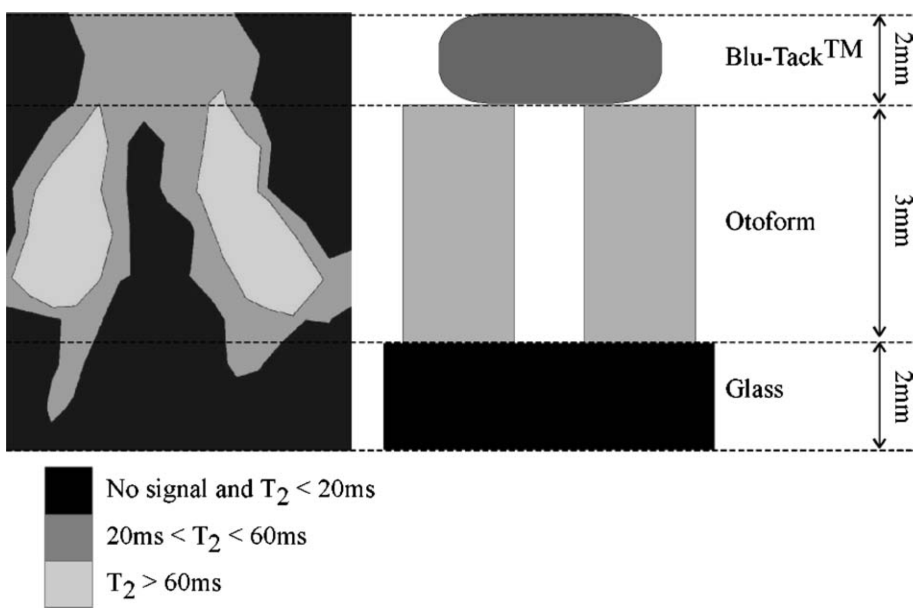

Fig. 3. T_2 contrast imaging of a Blu-Tack, Otoform, and glass phantom. The image plane is identical to that shown in Fig. 2C. The image is distorted to the irregular shape of the resonant volume. However, the three regions are distinguishable. The glass and air, containing no protons, have been shaded black. The Blu-Tack, with a short T_2 is dark grey, and the Otoform with a long T_2 is light grey. The edges of the Otoform pillars have become distorted and cannot be distinguished from the relaxation time of the Blu-Tack since it was difficult to fit a single exponential decay to the signal in partially filled voxels.

basic, low-cost, hand-held NMR sensor. Work has been presented elsewhere on a portable (but not palm-sized) three-dimensional MRI device [13]. Significantly the MRI-MOUSE design demonstrated here has the advantage of being small enough to be carried and operated by one person. The novel implementation of a sweep coil system [6] in a portable device has removed the need to manually retune the probe [5,14] to obtain spatial resolution vertically above the RF coil. One-dimensional profiling has been successfully achieved, and preliminary results on several multi-layer rubber and glass phantoms demonstrated a resolution better than 2 mm, and a sample penetration depth of 7 mm. For the future, the performance of the magnetic field sweep coil may be improved to reduce the required current and hence provide a more economic and portable set-up. It has been suggested that a car battery combined with a suitable time and amplitude regulator could be used to drive the sweep coil, making the system truly mobile. However, the rise and fall times of the sweep coil current are dependent on the source and it is doubtful that car batteries could provide the same performance as a 1 kW audio amplifier. Careful consideration would have to be given to such issues if this system were to be eventually implemented.

Two-dimensional images were produced in a plane normal to the surface of the MOUSE by adding a set of phase encoding coils (see Fig. 1A) [12]. The phase encoding sequence could be improved in the future by the implementation of oscillating gradient currents [15]. This would allow small current sources, e.g., car batteries, to drive the field gradients instead of audio amplifiers, making the system truly portable. T_2 relaxation time image contrast was also achieved, potentially leading to medical applications [16] of the MRI-MOUSE. This method could one day lead to the successful construction of a commercial, hand-held, portable MRI device that could have many industrial applications, such as cross-sectional imaging of mechanical rubber products, e.g., car tyres and airsprings on railway carriage suspension systems.

Acknowledgments

The authors gratefully acknowledge the help provided by Mr. Nathan Routley and Prof. John Strange for his support. We thank EPSRC for providing funding for this project.

References

[1] S. Sharma, F. Casanova, W. Wache, A. Segre, B. Blümich, Analysis of historical porous building materials by the NMR-MOUSE(R), Magn. Reson. Imaging 21 (2003) 249–255.
[2] G. Eidmann, R. Savelsberg, P. Blümler, B. Blümich, The NMR MOUSE, a Mobile universial surface explorer, J. Magn. Reson. Ser. A 122 (1996) 104–109.
[3] S. Anferova, V. Anferov, M. Adams, P. Blümler, N. Routley, K. Hailu, K. Kupferschläger, M.J.D. Mallett, G. Schoeder, S. Sharma, B. Blümich, Construction of a NMR-MOUSE with short dead time, Concept Magn. Reson. 15 (2002) 15–25.
[4] F. Bălibanu, K. Hailu, R. Eymael, D.E. Demco, B. Blümich, NMR in inhomogeneous magnetic fields, J. Magn. Reson. 145 (2000) 246–258.
[5] P.J. Prado, Single-sided imaging sensor, Magn. Reson. Imaging 21 (2003) 397–400.
[6] M.J.D. Mallett, M.R. Halse, J.H. Strange, Stray field imaging by magnetic field sweep, J. Magn. Reson. 132 (1998) 246–258.
[7] M.C.A. Brown, 2-Dimensional Magnetic Resonance Imaging with a Novel Handheld Sensor, PhD Thesis, University of Kent, Canterbury, 2003.
[8] M.D. Hürlimann, D.D. Griffin, Spin dynamics of carr-purcell-meiboom-gill-like sequences in grossly inhomogeneous B0 and B1 fields and applications to NMR well logging, J. Magn. Reson. 143 (2000) 120–135.
[9] H.Y. Carr, E.M. Purcell, Effects of diffusion on free precession in NMR experiments, Phys. Rev. 94 (1954) 630–638.
[10] S. Meiboom, D. Gill, Modified spin-echo method for measuring nuclear relaxation times, Rev. Sci. Instrum. 29 (1958) 668–691.
[11] P.T. Callaghan, Principles of Nuclear Magnetic Resonance Microscopy, Oxford University Press, Oxford, 1993.
[12] P.J. Prado, B. Blümich, U. Schmitz, One-dimensional imaging with a palm-size probe, J. Magn. Reson. 144 (2000) 200–206.
[13] J. Perlo, F. Casanova, B. Blümich, in: 3D Multi-Echo Imaging in Highly Inhomogeneous Magnetic Fields, 7th International Conference on Magnetic Resonance Microscopy, Snowbird, Utah, USA, 2003.
[14] F. Casanova, B. Blümich, Two-dimensional imaging with a single-sided NMR probe, J. Magn. Reson. 163 (2003) 38–45.
[15] S.P. Cottrell, M.R. Halse, J.H. Strange, NMR imaging of solids using large oscillating field gradients, Meas. Sci. Technol. 1 (1990) 624–629.
[16] P. Mansfield, P.G. Morris, NMR Imaging in Biomedicine, Academic Press, New York, 1982.