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Temperature Mapping via Relaxation-Based Color MPI

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Abstract

n magnetic particle imaging (MPI), the relaxation behavior of the nanoparticles has been exploited to expand quantitative mapping capabilities of MPI to applications such as viscosity mapping and temperature mapping. We have previously proposed a technique called TAURUS to estimate the relaxation time constant directly from the MPI signal, and demonstrated its viscosity mapping capabilities via imaging experiments. In this work, we extend TAURUS to demonstrate its temperature mapping capability via 1D imaging experiments at two different temperatures.

I Introduction

The relaxation behavior of the magnetic nanoparticles has been utilized for color magnetic particle imaging (MPI) in applications such as catheter tracking and steering [1-3], viscosity mapping [4,5] or temperature mapping [6]. These color MPI applications have broadened the quantitative imaging capabilities of MPI. One potential application is cancer imaging through viscosity mapping with color MPI techniques, as previous studies have shown that cancerous tissues have higher cellular viscosity levels [7,8]. Treatment monitoring can also be achieved by incorporating hyperthermic procedures. However, in such cases, it is crucial to heat up the diseased tissue only and not damage the healthy tissue nearby. Even then, the problem is challenging, as there is a confounding effect between viscosity and temperature on the MPI signal [9], which must be taken care of for accurate mapping purposes.

We have previously demonstrated a relaxation-based color MPI technique called TAURUS (TAU estimation via Recovery of Underlying mirror Symmetry) and showed its viscosity mapping capability via imaging experiments [5]. We further investigated a proof-of-concept analysis of this technique to estimate temperature using a magnetic particle spectrometer setup [9]. In this work, we extend TAURUS to demonstrate its temperature mapping capability via 1D imaging experiments.

II Material and methods

II.I Theory

In x-space MPI, the relaxation effects are modeled with an exponential kernel as follows [10]:

$$s(t) = s_{ideal}(t) * \{\frac{1}{\tau} e^{-\frac{t}{\tau}} u(t)\} \quad (1)$$

where τ is the effective relaxation time constant, $s_{ideal}(t)$ is the adiabatic nanoparticle signal, $u(t)$ is the Heaviside step function, and $*$ denotes convolution. The TAURUS technique estimates τ directly from the MPI signal, by using the underlying mirror symmetry of the adiabatic

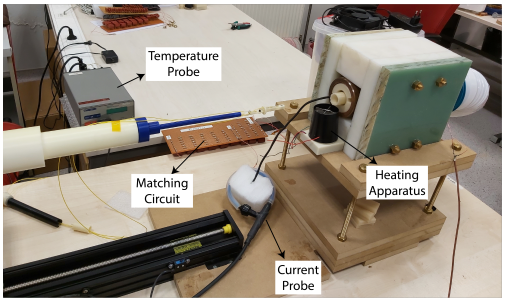

Figure 1: Our in-house MPI scanner. DF was applied at 15 mT peak and 9.8 kHz. The two samples on the imaging phantom were heated up using a custom-made apparatus.

signal. It should be emphasized that the effective τ does not directly correspond to Brownian or Néel time constants as they are modeled under zero-field. TAURUS can be directly applied to each partial field-of-view (pFOV) signal without any information about the nanoparticles as follows [11,12]:

$$\tau = \frac{S^*_{pos}(f) + S_{neg}(f)}{i2\pi f(S^*_{pos}(f) - S_{neg}(f))} \tag{2}$$

Here, $S_{pos}(f)$ and $S_{neg}(f)$ are the Fourier transforms of the positive and negative half cycles of the nanoparticle signal $s(t)$, respectively. The superscript * symbol denotes complex conjugation.

We have previously shown that different drive field (DF) frequencies and amplitudes can change the trends in τ. We have observed that operating around 10 kHz is well suited for τ mapping purposes for nanomag-MIP nanoparticles (which have the same chemical compound as the Perimag nanoparticles used in this work).

II.II Experimental Setup

Our in-house MPI scanner shown in Fig. 1 has two disc shaped magnets that generate a (4.8, 2.4, 2.4) T/m/μ_0 gradient in (x,y,z) directions, respectively. The drive field coil has 3 layers of 44 AWG Litz wire with 79 turns in each layer. The receive coil is a three section gradiometer type coil with a single layer of 40 AWG Litz wire. The middle and side sections have 34 turns and 19 turns, respectively. One side section of this receive coil can move independently from the other two sections and its position was adjusted manually to minimize the direct feed-through signal.

II.III Sample Preparation

Four different samples were placed with 1.8 cm separation in an imaging phantom with the same total volume of 20 μL, and the same nanoparticle amount of 10 μL of undiluted Perimag nanoparticles (Micromod GmbH, Germany) at 151.8 mmol Fe/L. These nanoparticles were

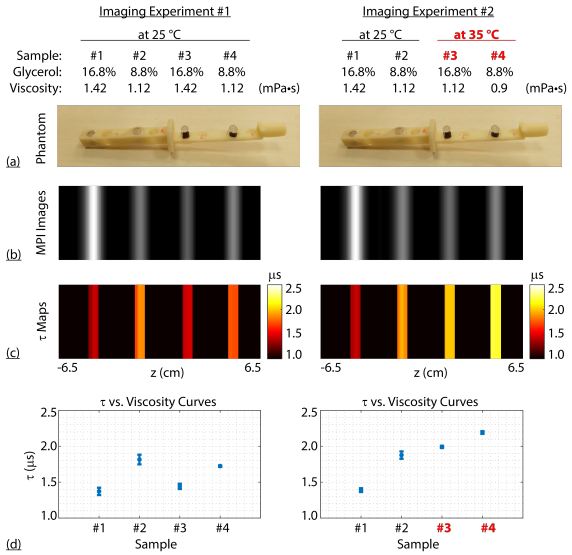

Figure 2: Imaging experiment results at 9.8 kHz and 15 mT-peak drive field. (a) The imaging phantom used at the two imaging experiments. (b) 1D MPI images of this phantom and (c) 1D τ maps, replicated in the vertical direction for display purposes. (d) τ vs. viscosity curves extracted from τ maps for each sample. Mean values and standard deviations were computed from the regions-of-interest of each sample.

mixed with different amounts of water/glycerol mixtures to obtain 8.8 % and 16.8 % glycerol percentage by volume. Two of each of these mixtures were placed consecutively on the imaging phantom as shown in Fig. 2(a). The viscosity values of these samples at 25 °C and 35 °C were interpolated using the values from [13].

II.IV Imaging Experiments

For 1D imaging experiments, a 13 cm field-of-view (FOV) was covered by 63 pFOVs with 85 % overlap between neighboring pFOVs. The drive field was applied in the z-direction at 9.8 kHz and 15 mT, corresponding to a pFOV size of 6.25 mm. Each pFOV signal was first low pass filtered with an analog Butterworth filter (Stanford Research Systems SIM965) at 300 kHz and then amplified with a voltage preamplifier (Stanford Research Systems SIM911).

The imaging phantom was separated with a plastic barrier so that the airflow was minimized between the two chambers at different temperatures - between the sample pairs (#1, #2) and (#3, #4). Two fiber optic temperature probes (Reflex-4, Neoptix) were placed in each chamber to monitor the temperatures throughout the experiments. The imaging phantom was taken out from the scanner via a linear actuator before the acquisition of each pFOV signal and the two samples were exposed to a convective heating at 60°C for 10-12 seconds. Then, the imaging phantom was placed back into the scanner

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and the measurement of the pFOV signal was performed once the targeted temperature of 35°C was reached.

As shown in Fig.2, two different imaging experiments were performed. In the first experiment, all four samples were kept at the same temperature at 25°C (left column). In the second experiment, two samples were at 25°C and the other two were at 35°C (right column).

III Results and discussion

The resulting MPI images, and τ maps of both experiments are shown in Fig.2(b) and (c), respectively. In Fig.2 (d), the τ values from these maps were extracted for each sample.

In the first experiment, sample pairs (#1, #3) and (#2, #4) had the same viscosity levels at 25°C and yielded approximately the same τ values, showing consistency of TAURUS estimations. In the second experiment, samples at higher temperatures have reduced viscosity levels. Accordingly, τ values of the samples (#3, #4) change due to the change in temperature. In fact, sample #3 at 35°C has the same viscosity level with the sample #2 at 25°C [13]. Hence, the changes in τ with temperature can be used to map temperature. As we have previously shown, τ values display different trends under different DF parameters. For a therapeutic application, imaging at two different DF parameters (by changing frequency or amplitude) can be performed to extract both viscosity and temperature maps.

IV Conclusions

In this work, we have demonstrated the capability of our relaxation-based color MPI technique, TAURUS, for separating different temperatures via imaging experiments. The results show promise for simultaneous mapping of viscosity and temperature, which will help expand quantitative mapping capabilities of MPI to cancer imaging and hyperthermic therapy monitoring applications.

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