Applying Magnetic Resonance Thermometry to Engineering Flows

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Abstract Measuring three-dimensional temperature fields simultaneously with three-dimensional three-component velocity fields by means of Magnetic Resonance Imaging (MRI) methods is the goal of a DFG-funded joint research project at Technische Universität Darmstadt and the University Medical Center Freiburg. This paper summarizes the basics needed to understand the principles of measuring temperature and velocity with MRI. A brief outline of former research applying MR Thermometry (MRTh) on engineering flows is given.

In a preliminary experiment temperature difference maps where achieved using MRTh for straight pipe flows at different constant fluid temperatures and different flow rates. Good agreement was found with simultaneously achieved temperature measurements using MR applicable fiber optical probes. A discussion on imaging errors occurring due to the experimental setup is given and important parameters influencing the MRTh sequence are described.

1. Introduction

MRI methods for the acquisition of flow quantities, such as velocity, concentration, turbulence and temperature have been established as a potential measurement technique for many fluid mechanical applications over the last years (Elkins and Alley, 2007). The combination of velocity fields and temperature fields are particularly interesting for the design and improvement of devices with heat transfer or heat exchange purposes. In the present research the challenge is to combine two different MRI methods: Magnetic Resonance Velocimetry (MRV) and Magnetic Resonance Thermometry (MRTh) in order to acquire spatially-resolved velocity and temperature data. Both MRI methods are based on a manipulation of the local Proton Resonance Frequency (PRF), which results in a temperature-induced phase angle change and a velocity-induced phase angle change of the complex MR signal. The phase angle contains both effects. The difficulty is to separate both effects from each other in order to not misinterpret the origin of the changes.

MR Thermometry is a medical tool for monitoring tissue temperature during minimally invasive thermal therapies, such as hyperthermia (43-45°C, treatment of cancer cells) or high-temperature thermal ablation (50-80°C, induce cell necrosis). The advantage of MRI-guiding during these medical therapies is to map the local tissue temperature with spatial resolution and in real-time. The most advanced and promising technique is PRF Thermometry, which has a minimal resolvable temperature of slightly less than 1K. However, until now MRTh has not become a conventional instrument for medical diagnostics. The acceptance of MRTh still suffers from non-reliable temperature measurements in non-aqueous tissue, such as fat, as well as artifacts due to patient movement or heat-induced tissue mutation (Rieke et al., 2008). These disadvantages are not present in the environment of a stationary flow model.
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Tab. 1: Comparison of MRTh with different temperature measurement techniques based on the table from Childs et al. (2000). The table shows the method name, the physical effect (if not defined by method name), the temperature range, the accuracy (Accu.), the response (Resp.), the measurement dimension (Meas. Dim.), the MR suitability (MR suit.), the costs and the commercial availability (Comm. avail.).

In feasibility studies researchers applied MRTh to generic fluid mechanical test cases. Elkins et al. (2004) used PRF Thermometry and measured a homogeneous temperature field of a turbulent pipe flow with constant fluid temperature.
With the use of a MRI machine with a very strong magnetic field (9.4 Tesla) Small et al. (2009) measured the temperature field inside miniature Shape Memory Polymer (SMP) foam, fed with a constant water flow rate. They also used PRF Thermometry. The foam was locally heated using a laser with a 7 Watt power maximum.

One of the most valuable studies in the field of MRTh was presented by Sun and Hall (2001). They considered the Graetz problem, which is fundamental in heat exchanger theory. They measured the flow and the temperature distribution inside a tube with constant flow and an applied temperature gradient along the streamwise direction. The flow enters the tube with a water temperature of 40°C and is then cooled down by water at 2°C flowing through a jacket surrounding the tube walls. In their study PRF Thermometry is applied simultaneously with MRV within one measurement procedure.

In comparison to other temperature measurement techniques, MRTh has certain advantages and disadvantages. Childs et al. (2000) review different temperature measurement techniques. Based on their work, Tab. 1 summarizes the features of different techniques, in order to compare with MRTh. Especially in cases of accuracy and measurement dimension MRTh is competitive to standard techniques, whereas the temperature range, the restriction to use water as measurement fluid, the missing transient capability and the response time summarize MRTh’s weaker points. Compared to other water-specialized temperature measurement techniques MRTh is equivalent or even superior. Tab. 1 also shows which measurement techniques can be applied within the MR environment for calibration and verification of MRTh measurements. Most applicable are fiber optical sensors, which are not influenced by the scanners magnetic field (thermographic phosphor) or can easily be calibrated (gallium arsenide crystals) for the use inside the MR scanner (Buchenberg et al., 2014). Another advantage is that different systems are commercially available. Other techniques, such as non-metal thermocouples or water temperature sensitive paint are still state-of-the-art and not commercially available.

All MRI methods have restrictions concerning the harsh magnetic and electromagnetic environment, the spatial resolution and the available space inside the MRI scanner bore. That makes it difficult to apply large enough temperature differences inside an MR compatible flow model of engineering relevance. The further development of MRTh methods for engineering purposes is justified, if appropriate experimental setups with adequate temperature gradients can be developed. Nevertheless, being able to measure full three dimensional three component (3D3C) velocity data as well as full 3D temperature data justifies such an effort.

2. MR Basics

The basis for MRI is the nuclear magnetic resonance (NMR) of atomic nuclei with uneven mass number. The human body mainly consists of water, a molecule with a combination of one oxygen and two hydrogen atoms. For this reason, MRI is generally tuned for hydrogen nuclei. Every atomic particle has a quantum mechanical spin, often denoted as a rotating impulse. This is the reason, why a spinning proton in an external magnetic field has a magnetic dipole moment, which, in turn, can interact with additional magnetic or electromagnetic fields. Inside an MRI scanner a strong static magnetic field $B_0$ is present, generated by a superconducting magnet. Due to the magnetic dipole moment, the hydrogen protons, when exposed to the main magnetic field $B_0$, start to precess around the magnetic field lines with the proton resonance frequency (PRF) $\omega_0$ (cp. Fig 1, left). The PRF is proportional to the gyro magnetic constant $\gamma$ and the applied magnetic field strength $B_0$, and is also known as the Larmor frequency.
Figure 1 Precessing proton around B-field with resulting PRF $\omega_0$ (left) and development of equilibrium magnetization $M_r$ (right).

Moreover they align along the magnetic field lines in two different quantum mechanical states with different energy levels: in a higher energy state, called down-spin state (anti-aligned with $B$) and in a lower energy state, called up-spin state (aligned with $B$). Both states are stable, but can be swapped from up-spin to down-spin by stimulating the protons with a certain amount of energy. The magnetic moments of one up-spin and one down-spin cancel each other. The difference in energy between the states leads to a slight excess of spins in the up-spin state, aligned with the magnetic field lines. By summation of all excessing up-spin dipole moments a longitudinal (aligned with the magnetic field) “equilibrium” magnetization $M_0$ is achieved (cp. Fig. 1, right). In a macroscopic sample of matter (e.g. a drop of water) the equilibrium magnetization is proportional to the magnetic susceptibility $\chi$ and the applied magnetic field strength $B_0$ but is about 6 orders smaller than $B$ (Haacke et al., 1999). Therefore, the equilibrium magnetization cannot be measured directly.

By stimulating the spins with a short radio frequency pulse (RF pulse or RF excitation), $M_c$ can be tilted into a plane perpendicular to the main magnetic field (transverse plane). The RF pulse consists of a linear polarized magnetic field rotating at Larmor frequency ($B$-field) and is provided by a transmitter coil (McRobbie et al., 2002). While the RF pulse is applied, the magnetization component along the B-axis ($M_z$) decreases and a transversal magnetization $M_x$ forms, also rotating at the Larmor frequency (cp. Fig 2).

Figure 2 RF pulse (B1-field) tilts the equilibrium magnetization $M_r$ into the transverse plane. The longitudinal magnetization component $M_z$ decreases, while the transverse magnetization $M_x$ forms.

After the RF pulse is switched off again, the spin system returns to the initial state. Thereby two temporally independent relaxation mechanisms occur. The magnetization $M_c$ regrows, until it equals the equilibrium magnetization $M_0$ again. This is termed longitudinal magnetization relaxation. The second mechanism is the decay of the transversal magnetization component $M_x$, which happens much faster than the first process. Due to the fact that $M_x$ rotates at the Larmor frequency, a coil sensitive for magnetization changes in the $x$-$y$-plane receives an induced decaying voltage, the free induction decay (FID). This is the MR signal. As $M_x$ is a complex quantity, it consists of a signal magnitude $|M_x|$ and a signal phase $\Phi$. The time constant between the transmission of the RF pulse and the receiving of the MR signal is known as the echo time TE. Those three measures are important for further considerations.

The difference between MRI and MR spectrometry is that MRI data has a spatial resolution. The MR signal arrives at the receiver coil as a superposition of all signals generated in the field-of-view (FOV). In order to accomplish spatial resolution, the FOV has to be partitioned into small discrete
volumes (voxel) and the MR signal coming from every voxel has to be coded uniquely. This complex procedure, called spatial encoding, is achieved by using the magnetic gradient unit of the MRI system. Each gradient coil generates a linear magnetic gradient along one spatial direction, its field strength adding to the main magnetic field B. This leads to a local manipulation of the B-field; hence, affects the local Larmor frequency. This local manipulation of resonance frequency has the effect that the signal received after a series of gradients is related to a single spatial frequency: the amount of signal received is proportional to the amount of that spatial frequency that is present within the object being imaged. The object is therefore imagined, not in position space, but in spatial frequency space (also known as k-space) and by applying many different gradients, many different spatial frequencies (k-space positions) can be measured, and an approximation to the object can be built up.

The following discussion assumes 2D imaging but the extension to 3D is simple. K-space (defined by two perpendicular axes, k and k.) is normally sampled by moving in a series of lines with constant k (this is known as Cartesian encoding). A short gradient along k ‘moves’ us to a particular k-space position (i.e. a particular spatial frequency in the y-direction) and is then switched off. This gradient is known as the phase encoding gradient and k, is known as the phase encoding direction. A constant gradient is then applied along k ‘moving’ us across a line of k-space. This constant gradient is known as the readout gradient and k is therefore known as the readout direction. While the readout gradient is being applied, the received signal is sampled at regular time points, hence a series of spatial frequencies with the same k, component but different components is measured. The process is repeated with many different k, values until a grid of measured points has been built up (the extent of the grid in k-space determines the final image resolution). From the fully sampled k-space, an image of the object can be recovered by performing a two dimensional Fourier transformation. To access three-dimensional imaging, modern procedures apply a second phase encoding gradient along the third perpendicular direction.

2.1 MRV

Measuring flows using MRV originated in medical diagnostics of the cardiovascular system of humans or animals. Up to the present, a variety of fluid mechanical flows have been measured using this measurement technique. One of the first applications was presented by Elkins et al. (2003), measuring the flow inside a turbine blade cooling channel with ribbed walls. Another study showed the highly three-dimensional swirling flow inside a tube with tangential inlets (Grundmann et al., 2012). Wassermann et al. (2013) analyzed the unsteady flow inside a bi-stable fluidic oscillator by using phase-locked 3D3C MRV.

Magnetic Resonance Velocimetry is an application of MRI, utilizing the sensitivity of the signal phase (phase contrast imaging) for fluid flow (Bryant et al, 1984 or Markl et al., 2003). Based on the gradient-echo sequence a time-averaged velocity field is achieved over multiple repetitions. The target of flow encoding is to add a phase offset to the measured magnetization which is proportional to the velocity of flowing spins within each voxel. This is done with the help of a bipolar gradient. Assuming that spins are moving constantly in the same direction as a magnetic gradient (amplitude +Gk), they acquire phase as shown in Fig. 3. After duration Δt the amplitude of the gradient is reversed (-Gk). During the reversed gradient duration flowing spins lose more phase than they gained. After the bipolar gradient, each Voxel has acquired a velocity-dependent phase shift Φ. In contrast, static spins gain and lose the same amount of phase and so are uninfluenced by flow encoding. However, in addition to the phase caused by the velocity of the spins, other effects (such as inhomogeneities in B.), contribute to the phase. If the measurement is repeated with an equal gradient of opposite polarity, the two phase shifts in each voxel can be subtracted. The resulting phase difference ΔΦ is proportional to the fluid velocity only. At this point it is recommendable to introduce the velocity encoding value Venc. It describes the maximum resolvable fluid velocity at which the phase difference ΔΦ = π. The Venc is a control parameter combining the adjustment of gradient amplitude Gk and gradient duration Δt.

In modern MRV sequences a modified bipolar gradient, also known as velocity compensation gradient (cf. Fig. 3 middle), is applied and subtracted with a flow encoded data set per direction.
Velocity compensation only works fully for steady flow conditions. This gradient is also applied in MRTh.

Another feature is the possibility to measure unsteady cyclic flows using phase-locked or triggered MRV (4D-MRV). Hereby, the MR scanner is triggered on the beginning of each flow cycle (e.g. heart flow is triggered by human ECG). The cycle is then subdivided into phase angles, which give the pseudo-time resolution. The MRV data set is filled with every measured flow cycle, so that the flow field of every phase angle is composed of information from different flow cycles.

![Fig. 3](image)

**Fig. 3** Left: Flow encoding gradient (yellow) shows the reaction on the signal phase of static and moving spins. Middle: Flow compensation gradient (yellow) produces a zero net phase of static and moving spins. Right: Sequence diagram for a 3D3C MRV acquisition. RF is the radio frequency excitation; $G_s/G_r/G_e$ is the sequence of all gradients this direction; Signal is the timeline showing the duration until the signal echo arrives (TE) and the duration until the next RF repetition (TR). The bipolar gradients achieve flow encoding.

The blue and red (dashed) gradients are the phase encoding steps, in which the gradient amplitude is transposed every TR. Green is the frequency encoding gradient, which evokes the formation of the signal echo.

### 2.2 MRTh

There are different ways of extracting temperature information from the various MR quantities. A very promising one is PRF Thermometry, which uses the temperature dependent properties of water molecules ranging from liquid to crystal state. In order to understand the link between magnetic and temperature-dependent properties of water molecules an appropriate understanding of hydrogen bonding is necessary. In water hydrogen bonds develop due to intramolecular dipole-dipole interactions between the proton donor ($\delta^+$, hydrogen) and the proton acceptor ($\delta^-$, oxygen), as depicted in Fig. 4. The larger the bond angle $\Theta$ and the shorter the bond length is, the higher is the bond energy and, consequently, the bond strength (Jeffrey, 1997). The hydrogen bonding parameters of water molecules are temperature dependent. With increasing water temperature the hydrogen bonds bend ($\Theta$ decreases), stretch and finally break. This process leads to a rearrangement of the water molecules’ electron cloud, which, in the hydrogen bonded state, is displaced from the molecule’s electric center. The electron cloud now has the ability to shield the hydrogen core more efficiently (Hindman, 1966).

![Fig. 4](image)

**Fig. 4** Water molecules with hydrogen bond and bond angle $\Theta$
Increasing or decreasing fluid temperature changes the shielding constant of the 1H protons linearly with the proportionality ratio \( \sigma(T) = \alpha \cdot T \) in the temperature range from \( T = -15 \) to 100°C with a proportionality rate of \( \alpha = 0.01 \text{ppm} \left( ^\circ \text{C} \right)^{-1} \) (Hindman, 1966). This value is also called screening constant and was experimentally derived for pure water. The exact value for a different fluid should be determined prior to any MRT measurement. Increasing or decreasing fluid temperature affects the local effective magnetic field \( B_0 \) and, thus, affects the effective PRF \( \omega_{\text{eff}} \) as given in Eq. 1.

\[
\omega_{\text{eff}} = \frac{\gamma B_0(1 - \sigma(T))}{B_{\text{eff}}}
\]

Consequently, the temperature-dependence of \( B_0 \) leads to a temperature-dependence of the PRF. For a given echo time TE, and according to the actual fluid temperature, which is assumed steady during TE, a certain MR signal phase angle \( \Phi \) is acquired. As mentioned above, a velocity compensation gradient is included to remove any dependence of the phase on velocity. Two scans at different temperatures are required to produce a temperature difference map. Firstly a reference scan is acquired at a spatially constant fluid temperature \( T_\text{ref} \). Then, the same MR sequence is applied on a case with a different temperature distribution to the reference scan, \( T \) (this can be a spatially constant higher or lower temperature or a spatial temperature distribution). By subtraction of the two MR signal phase angle distributions of the scan with temperature field applied (heat on) \( \Phi(T_\text{ref}) \) and the reference scan (heat off) \( \Phi_{\text{ref}}(T_{\text{ref}} = \text{const}) \) the temperature-dependent phase lag can be calculated (Ishihara et al., 1995):

\[
\Delta \Phi = \Phi_{\text{ref}}(T_{\text{ref}}) - \Phi(T_\text{ref})
\]

The temperature difference map can be calculated from the phase difference image (\( \Delta \Phi \)) and the given MR parameter (Rieke et al., 2008):

\[
\Delta T = \frac{\Delta \Phi}{\alpha_T \text{TE}_0}
\]

If \( T_\text{ref} \) is known, for example, by a simultaneous measurement of the reference temperature with a MR-suitable temperature sensor (fiber optics) during the reference scan, the absolute temperature field can be estimated, by addition of \( T_{\text{ref}} \) and \( \Delta T \).

3. Experimental Setup

3.1 Flow Supply System and Flow Instrumentation

A flow supply system, placed outside the MR scanner room, provides the water flow. It consists of a pump, a flow meter, a pressure transducer and a temperature sensor. All components are either controlled or monitored using a Labview data acquisition system. The conditioned flow is fed through PVC supply hoses to the test section. Connected to the test section’s outlet is a draining hose, leading the water back to a cart with a 100L tank. The fluidic circuit is closed via a connection hose between tank and suction side of the pump. A water conditioning system keeps the water temperature inside the tank at a fixed value. It consists of an electrical heating foil with polyimide insulation (Minco HK5600) and a Julabo FT402 immersion cooler. Fig. 5 shows the flow supply system and its components.

An optional device can be installed which provides an adjustable supply voltage for in-situ placed heating elements. The heater control system is connected to the Labview system and is capable of operating the heating elements cyclicly. Additionally, a 5V-TTL trigger is sent to the MR scanner marking the beginning of each new heating cycle. Triggering enables phase-locked MRV.

For validation, in-situ temperature logging is done with an Optocon® Fotemp 4-channel fiber optical thermometer. The probe consists of a glass fiber with a gallium arsenide crystal at the tip, which utilizes the temperature-dependent position of the band edge. Prior to a measurement the
probe has to be calibrated inside the MR scanners’ magnetic field (Buchenberg et al., 2014).

Fig. 5 Schematic and photograph taken from the flow supply system in the support room. The tank cart (light blue) is visible in the foreground with the immersion cooler (magenta) adjusted on the left side. Following is the pumping cart (green) with a Labview system (dark blue) on top, controlled by a Laptop computer (yellow) standing on the desk. Additionally the Labview system controls the optional heater control system (orange), which provides the supply voltage to the heating elements inside the flow model. The trigger signal for the triggered MR sequence is also provided by the Labview system. A fiber optical temperature sensor system controlled by a second Laptop is capable of measuring and logging fluid temperatures inside the flow model. The hoses and cables are put through special holes (red) in the wall, preventing irradiation to enter the MR scanner room.

3.2 Flow Model

As a first setup for MRTh measurements a straight pipe model was developed (cp. Fig. 6). A diffuser decelerates the fluid from the hose diameter of 25mm ID to the pipe diameter of 50mm ID. Three internal grids help the flow to prevent separation. The pipe has a total length of 350.7mm and is made of acrylic glass. The nozzle leads the fluid back into the hose connected to the flow supply system. Between the pipe and the diffuser/nozzle is an adapter part, which has holes for temperature probes. All parts, except the pipe are made of polyamide using direct laser sintering. The parts are mated with PVC screws and sealed with o-rings.

Fig. 6 Setup of the pipe model.

3.3 MR System and Procedure
The MR machine used is a Siemens Magnetom Trio with a 3 Tesla main magnetic field. The device is in property of the University Medical Center Freiburg and is operated by trained personnel from the Department of Radiology, Medical Physics. A special software interface enables the researcher to manipulate the measurement sequence parameters. The scanner stands in a separate room within a Faraday cage, which prevents electromagnetic waves from entering and producing imaging artifacts. The patient table holds the flow model (connected to the hoses and standing on a drip tray) and can be moved into the bore. Additionally, an RF coil is installed underneath the bolster. In order to gain sufficient SNR another coil (typically a body array coil) is adjusted above the measurement section. When the desired measurement section exceeds the range of the covering coil an additional one can be applied. A typical setup is illustrated in Fig. 7. Just before the measurement begins, the patient table is automatically moved, placing the measurement section into the iso-center of the magnet.

MRV experiments are time-consuming preliminary to the measurements, whereas data acquisition and data processing is a rather fast process (within minutes). Typically, in one measurement campaign (1 day ~ 8h) different flow models and/or different flow parameters of the same model are measured.

![Fig. 7 MR scanner (light blue), bore (dark blue), patient table (red), flow model on drip tray (green) and coil (yellow) put on top of the measurement section.](image)

There are many imaging parameters in an MRTh sequence which can be adjusted to optimize a particular measurement. All of these need to be considered for each individual flow and temperature distribution. For thermometry measurements the echo time TE is particularly important: as can be seen in Eq. 3, the amount of phase change for a given temperature change is most effectively controlled by changing TE. The greater the phase change for a given temperature change, the less noisy the measurement becomes, indicating that TE should be chosen as large as possible. However, in practice this is not possible. Firstly, analogous to the Venc in MRV measurements, there is a maximum temperature change that can be measured for a given TE which corresponds to the temperature at which the phase change is $\Delta \Phi = \pi$. Secondly, errors due to field inhomogeneities (which could be significant in a complicated setup consisting of parts constructed out of different materials with different susceptibilities) increase with increasing TE. The echo time is therefore set to be as short as possible while still yielding sufficient SNR for the range of temperatures expected. For reference, with pure water and at field strength of 3 Tesla, the maximum resolvable temperature difference (Kelvin) is approximately 400 K ms / TE [in ms].

Other parameters to be considered are the image resolution and the FOV size. To increase image resolution more data must be collected and hence the acquisition time must be increased (doubling resolution (in 2D) typically doubles acquisition time). Meanwhile, the FOV must be large enough for the object to fit within the phase encoding directions. Otherwise aliasing occurs, leading to wrapping artifacts.

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<th>2D/3D Triggered?</th>
<th>FOV [mm]</th>
<th>TE [ms]</th>
<th>Spatial resolution</th>
<th>Time per line</th>
<th>Lines per phase</th>
<th>Temporal resolution [ms]</th>
<th>Relative SNR</th>
<th>Scan time [h:m:s]</th>
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Another complication to be considered is triggering the MRTh measurements for acquiring phase averaged velocity/thermometry data. This is important in setups within which the temperature distribution is caused by Ohmic heating: flowing current within the object causes a large amount of artifact, effectively destroying the resulting MR images. To create an inhomogeneous heat distribution electric heating must therefore be cyclic (for example with heater on for 0.5 seconds and then measuring with heater off for 0.5 seconds), and the imaging must be triggered to the same cycle. In this way a heat distribution can be created and maintained but data can be acquired which is free from the detrimental heating artifacts. When triggering is used an additional trade-off between scan time and temporal resolution is created. A large amount of data for each phase angle (a high number of k-space lines per phase angle) can be acquired in each cycle, leading to a short scan time. However the temporal resolution will be low. Acquiring less data in each cycle (fewer k-space lines per phase angle) leads to more phase angles and, hence, better temporal resolution, but increases acquisition time.

Tab. 2 provides some typical scan times for various combinations of parameters (triggered acquisition times assume a one second cycle).

### 4. Preliminary Results

An initial experiment has been performed to acquire MRTh measurements in flowing liquid and to demonstrate some of the difficulties that can be faced when attempting to set up and measure temperature distributions using MRTh. The equipment was constructed as described in section 3, with a cylindrical section in the region of interest to allow a simple flow profile. The liquid used was distilled water with added copper sulfate to increase MR contrast. The addition of copper sulfate can affect the screening constant $\alpha$ and so $\alpha$ was determined experimentally in a separate experiment similar to that described in Peters et al. (1998). The water was heated to several different temperatures and at each temperature 2D MRTh measurements were carried out in a plane perpendicular to the flow direction (cp. Fig. 6). Several fiber optical probes were placed inside the flow and within the reference phantoms to validate the MRTh measurements and to measure any background changes in temperature during the experiment. Reference phantoms were placed in the field of view to allow correction for field drift and to demonstrate some of the errors that can arise if the experiment is not properly designed. A magnitude MR image of the setup is shown in Fig. 8.
Fig. 8 Experimental setup. The tube of heated liquid (circular object) was surrounded with reference phantoms to allow for a background correction. The red ROI represents the area averaged for temperature measurements. The green ROIs represent the areas used for background phase fitting and subtraction.

Ideally reference phantoms should surround the region of interest on all sides to allow an accurate background fit. However this must be carefully considered: the positioning of the reference phantoms in this case causes several problems due to susceptibility artifacts. Every object inside a homogeneous magnetic field causes a change in the local B-field. Especially at material boundaries (water-air, water-plastic, air-plastic) different magnetization potential (susceptibility) leads to B-field inhomogeneities and, as a consequence, to measurement artifacts.

Fig. 9 shows an example phase map subtraction for two acquisitions at different temperatures with a long echo time of 20ms. The long echo time highlights artifacts caused by susceptibility differences between materials which appear as false temperature difference distributions. The main problems arise due to the phantoms being too close to each other and thereby causing susceptibility artifacts, both in the phantoms themselves (making an accurate background correction difficult to achieve), and in the region of interest (causing inaccurate temperature measurements). Additional artifacts can be seen due to the presence of the temperature probe and air bubbles at the top of the pipe. The regions of interest for the background correction and for the temperature measurements were therefore chosen to lie in areas unaffected by artifacts (the regions used are marked in Fig. 8).

Furthermore, Fig. 9 demonstrates the necessity of background phase subtraction. A phase slope can be seen in the large reference phantom on the right, due to a B-field drift between the two subsequent measurements used for subtraction. Without having the reference phantoms within the FOV, this field drift would contribute to the temperature measurements and cause errors. With the reference phantoms present, the offset caused by the drift can be modelled and finally corrected for.

Fig. 9 Example temperature map with a long echo time of 20ms. Field inhomogeneities, due to locally varying magnetization potential (susceptibility) cause errors in the temperature maps. They need to be
Additional measurements were carried out with an echo time of 10ms (leading to a measureable temperature difference range of approximately ±40K). In sum 12 scans were carried out: the reference and three spatially constant fluid temperatures were achieved (room temperature, +5K, +10K, +15K). Each temperature case was measured at three different flow rates (20, 40 and 60 l/min). The three scans achieved at room temperature were used as reference scans for phase subtraction to produce temperature difference maps. A background phase correction was performed to remove the phase contribution due to field drift between the acquisitions. The average temperature in a region within the tube was calculated and compared to the optical probe values.

**Fig. 10** shows the results of the performed experiment. Temperature changes measured with MRTh are close to the values measured by the optical thermometer (within 1.2K for all measurements). Apparently, they are unaffected by variation of the flow rate. However, the temperature distribution within the circular cross section is clearly inhomogeneous. Air bubbles at the top of the tube cause erroneous ‘hot spots’ and the reference phantom to the left of the tube causes the measured temperature difference on that side of the tube to be artificially raised. A readjustment of the flow model and reference phantoms inside the MR scanner could avoid these artifacts.

The values measured by MRTh slightly underestimate the temperature change as measured by the optical probes. This could be caused by the susceptibility artifacts impeding an accurate background phase correction. The temperature underestimation could also be due to errors in the experimental determination of the proportionality rate $\alpha$, which is used to convert phase difference maps into temperature difference maps. However, the experiment confirms that the described MRTh procedure is capable of measuring temperature changes in flowing liquids.

**Fig. 10** Temperature maps in the pipe with heated fluid at three different temperatures and three different flow rates. The temperatures measured with optical probes are shown above each column. The number within each image is the mean and standard deviation of the temperature change within the ROI marked in red in **Fig. 8**. The echo time was TE=10ms for all cases.
5. Conclusion and Outlook

This paper describes the efforts towards a procedure for measuring the 3D temperature field of the fluid in a complex geometry together with the three dimensional three component velocity field. In order to elaborate the requirements and limits of this measurement technique a brief overview of the basic principles for measuring temperatures using a magnetic resonance scanner is given. Then the experimental setup that was developed for this project is described in detail and a description of the data acquisition is given.

The results of a first set of experiments conducted in a generic pipe flow are discussed and the current state-of-the-art of this measurement technique is demonstrated. So far it is possible to measure a homogeneous temperature field at different temperature levels and different flow rates without an influence of the flow on the temperature measurement. Difficulties arise from changes of the susceptibility of different materials within the FOV. Different susceptibilities lead to artifacts in the measured image that need to be compensated for. Procedures to compensate for such artifacts or to avoid them will have to be developed.

The next step of this project is the measurement of inhomogeneous temperature distributions in flows. The first challenge on faces is the creation of temperature differences in a flow inside the MR scanner. Electrical heating elements made of ceramics (Bach Resistor GmbH) were used for in-situ heating. A backward-facing step model was designed containing four ceramic heating elements installed at the bottom after the step, so that the recirculating fluid with low local flow velocities passes the heating zone and heats up. With a heating power of less than 250W, temperature differences of up to 20K could be achieved. The preliminary MRTh measurements (triggered sequence, due to electric heating) showed large artifacts produced by the changing magnetic susceptibility of the hot ceramic elements. The phase error resulting from susceptibility artifacts was of the same order as the temperature-dependent PRF shift of the fluid and, hence, no temperature difference could be worked out. De Poorter (1995) presents a way to model the susceptibility effects of different tissue and geometry. A future approach could be to properly simulate the susceptibility-induced phase change of the ceramic material and then subtract the resulting phase map from the measured phase difference map. Among many other project steps, this looks to be a promising way of extracting temperature maps from existing data sets.

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