Assessment of Temperature Changes caused by 7T MRI Scans using Proton Resonance Shift Thermometry

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Abstract – Magnetic resonance (MR) thermometry is used to access the temperature of a subject during an MR imaging (MRI) procedure. Measuring temperature in MRI is notably important for analyzing the safety of the subject, for better control of an intentional or unintentional heating, for experimental validation of modelling results, among others, and it is especially relevant for MR applications in ultra-high fields (i.e. $B_0 \geq 7$T). There are different techniques used as MR thermometry, which are capable to acquire the subject's temperature locally and/or spatially. The proton resonance frequency (PRF) shift method presents high resolution, fast acquisition, non-invasiveness and relatively accurate results [1]. In this work, we used the PRF shift method to evaluate the temperature changes occurring in a muscle-mimicking phantom that was exposed to the radiofrequency (RF) heating applied by a birdcage head coil during a 7T MRI scanning. The temperature rise in the phantom's center was also measured using a digital thermometer. The results for both thermometer and PRF shift method (after the deduction of the B_0 drift effects) are comparable. They showed a temperature rise of the phantom by around 3°C after 112 minutes of applying a pulse sequence protocol, which was used to simultaneously heat the probe and acquire the phase information.

Methodology: The muscle-mimicking phantom is a water-gelatine-oil-salt mixture based in [2]. It was manufactured as described in [3], and the mixture was placed in a cylinder made of polymethyl methacrylate (PMMA, Plexiglas®, Germany) with 50 mm of radius and 200 mm of height (Fig. 1a). The phantom's dielectric properties were measured using a dielectric probe kit (DAK-12, SPEAG, Switzerland) equipment. Measurements for 300 MHz were repeated in 8 different regions of the phantom. The results for conductivity and permittivity ($\sigma_p = 0.742 \pm 0.006\,\mathrm{S/m};\ \varepsilon_p\ = 57.65 \pm 0.45$) presented good agreement with literature values [4] and verified the phantom's homogeneity.
The MR measurements were done at a 7T MRI scanner (MAGNETOM 7T, Siemens Healthcare, Germany). A pulse-sequence protocol involving a combination of gradient-echo (GRE) and turbo spin-echo (TSE) sequences was implemented to acquire phase information for PRF shift and to heat the phantom, respectively. The GRE phase images were acquired before, after and between the TSE applications. A better description of this process and the sequences parameters are presented in [3]. To calculate the temperature increase of the phantom between time points t_f and t_0, for $t_f > t_0$, the

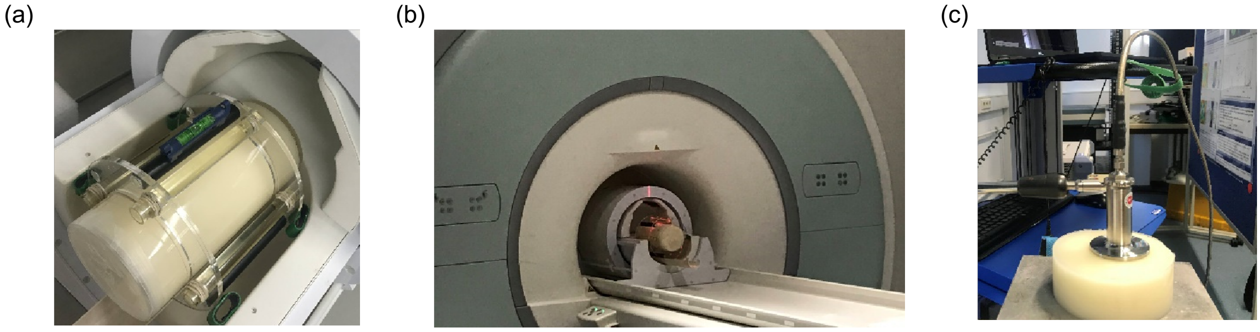

Fig.1: (a) Phantom and oil tubes inside the RF coil. (b) The measurement setup in the MR scanner. (c) Measuring the dielectric properties of the phantom mixture using the DAK equipment.

acquired GRE phases φ_0 and φ_f were used. The temperature difference ΔT_f was calculated using the PRF shift formula: $\Delta T_f = (\varphi_f - \varphi_0)/\alpha\gamma B_0 T_E$, for $\alpha = -0.01\ \mathrm{ppm/°C}$ (PRF coefficient), γ the gyromagnetic ratio of hydrogen and echo-time $T_E = \ 5.5\ \mathrm{ms}$. The phantom's phase variation comes mostly from the heating, but the fluctuations of the B_0 field during the measurement also counts for it. To dismiss the last one, four oil tubes ($h = 130\ \mathrm{mm};\ r = 12\ \mathrm{mm}$) were measured along with the phantom, because the oil's phase vary predominately due to the B_0 drift. Finally, phase correction maps covering the region of the phantom were calculated interpolating the oil's phase changes, and the phantom's temperature variation was calculated as $\Delta T_f = ((\varphi_f - \varphi_0) - \ (\varphi_f^{oil} - \ \varphi_0^{oil}))/\alpha\gamma B_0 T_E$.

Results: The PRF shift temperature rise maps for the central axial slice were calculated for $t_f = 18, 31, 44, 56, 69, 82, 95, 108\ \mathrm{min}$. They are presented in Fig. 2 with and without B_0 drift corrections.

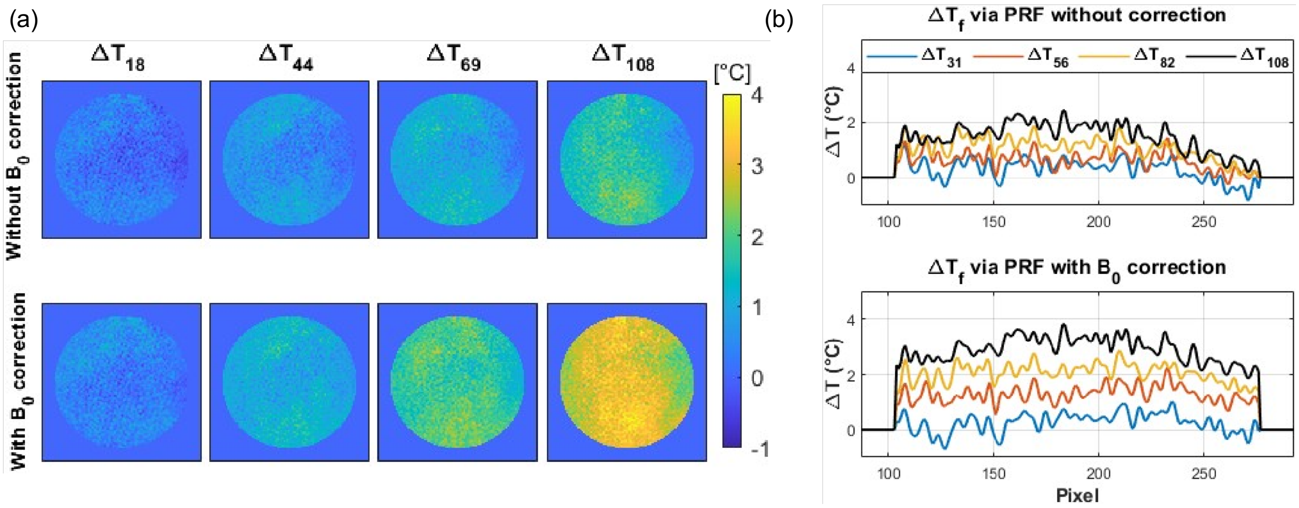

Fig.2: (a) Phantom's temperature maps obtained using the PRF shift method for the central axial slice. (b) The temperature variation along the horizontal line in the center of the phantom.

The central phantom's initial and final temperature was measured outside the MR room using the digital thermometer ($T_0 = 17.4 \pm 0.1\ °\mathrm{C}$ and $T_{112} = 20.5 \pm 0.1\ °\mathrm{C}$). Thermometer temperature values were linearly interpolated for the other t_f. Comparisons of thermometer and PRF shift method with B_0 drift corrections presented good accuracy, varying less than 0.5 °C.

Discussion and conclusion: This work used the PRF shift technique combined with thermometer measurements to asses the temperature rise of a muscle-mimicking phantom, which was exposed to RF heating applied by a birdcage head coil for 7T. This study presents limitations. Continuous temperature measurement, as well as a faster GRE acquisition could improve the accuracy of the technique. The PRF method used was successful to analyze the temperature rise spatiotemporally.

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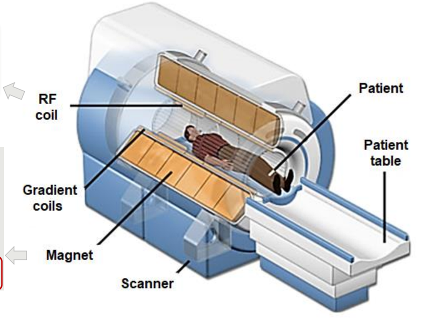
Motivation

Magnetic resonance imaging (MRI)

- Non-invasive imaging technique.
- Used for detection, diagnosis and treatment of diseases.

Transmission (Tx) and receiving (Rx) of signal

B_0 (T)	ω_0 (MHz)
1.5	63.87
3	127.74
7	298.06



Moore J. and Zouridakis G. Biomed. Tech. devices, CRC Press, USA. 2004.

- **Ultra-high field (UHF) problems:**
  - Electromagnetic fields are highly complex and spatially non-uniform.
  - The radio-frequency (RF) fields used induce tissue heating.

Safety in MRI

- Excessive heating of body tissues poses risk of subject injury.
- Safety guidelines (IEC 60601-2-33):
  - Temperature limits.
  - Specific absorption rate (SAR) limits.
- Heating and SAR correlation is non-linear and heterogeneous.

Safety

Analysis via simulations

Analysis via measurements

- **SAR:**

$$SAR = \frac{\sigma|E|^2}{2\rho}$$

ρ : density [kg/m³], σ : electrical conductivity [S/m].

- **Temperature:**

$$\rho c_p \frac{\partial T}{\partial t} = \nabla \cdot K\nabla T + q_m + \omega_b \rho_b c_b (T_a - T) + SAR$$

K : thermal conductivity, C_p : heat capacity, ω_b : perfusion, T_a : arterial blood temperature, q_m : metabolic heat source, SAR: volumetric heating rate generated from external sources.

- **SAR:** cannot be measured.
- **Temperature:** measured using MR thermometry.

MR Thermometry

Analysis via measurements

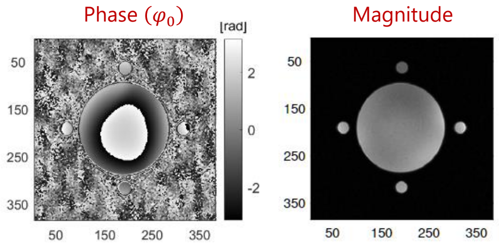
1. Locally, using temperature sensors.
2. Spatially, using the MR scanner as a tool.

Proton resonance frequency (PRF) shift

- Spatio-temporal evaluations, non-invasive, accurate results.
- Based on the fact that the electron shielding of water varies linearly with the temperature.
- Temperature variation is detected by evaluation of the phase change in the heated area.

$$\Delta T_f = \frac{\varphi_f - \varphi_0}{\alpha\gamma B_0 TE}$$

φ: phase measured, γ: gyromagnetic ratio of H, TE: echo time, B_0: main magnetic field, α: PRF thermal coefficient (α=-0.01ppm/°C).



- **PRF shift with B_0-drift corrections:**

- Field drift over time from the superconducting magnets are the major difficulty for long experiments (1 to 2 hours).

-> **Influence:** The field drift alone can account for a temperature error of ~2°C in a single hour.

-> **Correction:** Using external lipid reference probes such as oil. Lipid resonance frequencies show almost no temperature dependent shift. Thus, any change will be due to magnetic field drift.

$$\Delta T_{f_c} = \frac{(\varphi_f - \varphi_0) - (\varphi_{oil_f} - \varphi_{oil_0})}{\alpha\gamma B_0 TE}$$

Temperature measurement

Phantom preparation

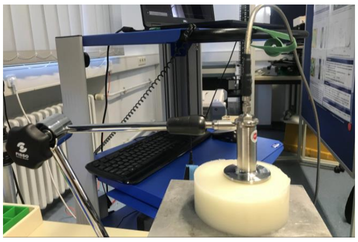
- Homogeneous muscle-mimicking phantom to simulate the dielectric, thermal, and MR properties of human muscle.
- Made of a water-gelatine-oil-salt mixture.

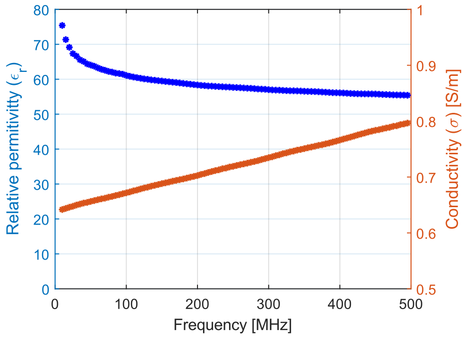
Yuan et al. Phys Med Biol 2012; 57(7):2021-37.

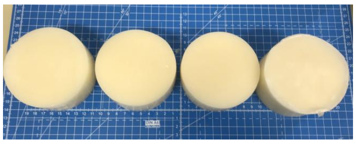
- Inserted into a cylindrical container of Plexiglas (h = 200mm, r = 50mm).

Phantom measurement

- Dielectric properties were measured using a dielectric assessment kit (DAK-12, SPEAG, Switzerland).
- Measured at 23 °C over the frequency range of 10 - 495 MHz, and repeated in 8 different regions.







- Results at 300 MHz :

$$\sigma_p = 0.742 \ \pm 0.006 \text{ S/m},$$

$$\varepsilon_{r\ (p)} = 57.65 \ \pm 0.45$$

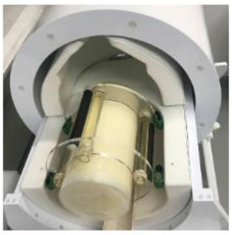
- Literature values:

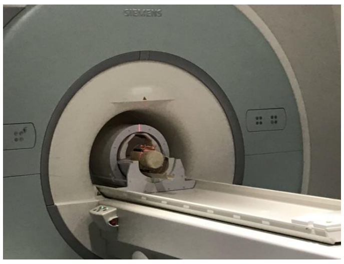
σ= 0.771 S/m and ε = 58.2.

Temperature measurement

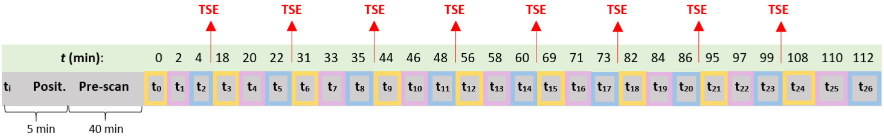
RF heating and temperature imaging

- 7T MRI scanner (MAGNETOM 7T, Siemens) installed in São Paulo (Brazil) and Magdeburg (Germany).
- Structure of Plexiglas to hold the phantom and oil tubes (l = 130mm, d = 12mm).
- RF coil: Birdcage head coil 1Tx/32Rx (Nova Medical).





- **RF heating and imaging protocol (PRF):**



- Gradient-echo (GRE) pulse sequence used to obtain phase information at different time points, and for all three anatomical planes (axial, sagittal, and coronal).
- T2-weighted turbo spin-echo (TSE) sequences to heat the phantom.
- A digital thermometer (TP101, Bmax) was used to control the temperature.
- Phantom rested overnight in the MRI room to ensure initial thermal equilibrium.

GRE parameters:

- TE = (2.5, 4, 5.5, 7) ms, TR = 10 ms, flip angle = 30°.
- FoV = 192 x 192 mm, matrix size = 384 x 384.
- N. slices: a = 20, c = s = 13, slice thickness = 5mm.
- Acquisition time: a = 11.0 s, c = s = 7.3 s.
- All phase-encoding directions and orientations.

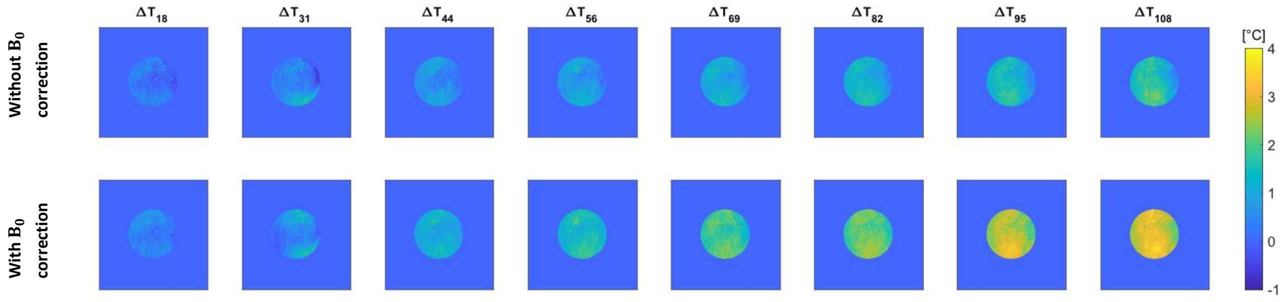
TSE parameters:

- TE = 24 ms, TR = 5 s, flip angle = 180°.
- FoV = 192 x 192 mm, matrix size = 192 x 192.
- N. slices: a = 13, slice thickness = 10 mm.
- Total acquisition time: a = 8.5 min.
- One phase-encoding direction.

Results

Results

- **Thermometer measurements at phantom's isocenter:**
  - $T_{room}\,(t_0) = T_p(t_0) = 17.4 \pm 0.1$ °C.
  - $T_p(t_{112}) = 20.5 \pm 0.1$ °C.
  - $T_p(t_{18})$ to $T_p(t_{108})$ were interpolated.
- **Temperature maps calculated via PRF shift method with and without B_0-drift correction (central axial slice, TE =5.5 ms, AP):**

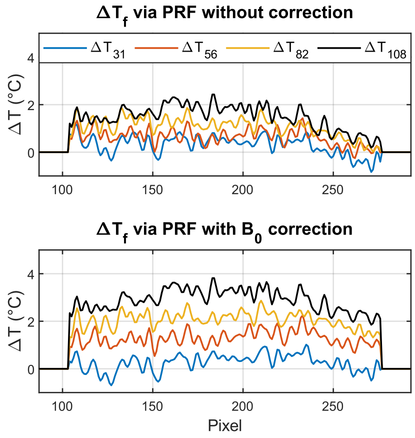


  - Temperature increase in time and space can be seen for both cases.

Results

Results

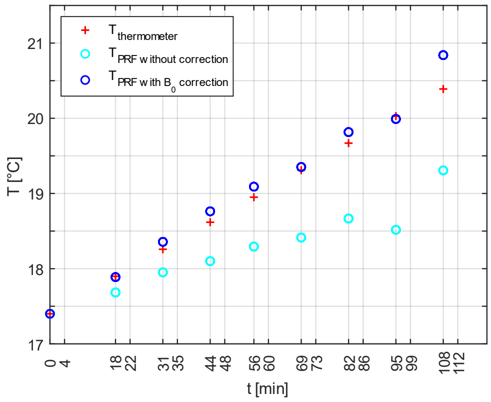
- **Temperature variation along the central horizontal line in axial view:**
  - Curves plotted from raw data.



- Temperature variations in time and space increased after B_0-drift correction.

Results

- **Phantom's temperature at isocenter**
  - PRF shift: calibration $T_{PRF}(t_0) = T_p(t_0)$, and ROI = 3 pixels-radius.



- The PRF method present an accuracy increase after correcting the B_0-drift effects using the oil tubes.

Conclusions and outlook

- The created muscle-mimicking phantom presented dielectric characteristics comparable to muscle tissue.
- RF heating protocol, applied by the birdcage head coil for 7T, elevated the phantom's temperature by around 3°C.
- PRF shift method provide time and spatial information from the RF absorption during 7T MRI.
- Temperature variation measured via PRF shift presented an accuracy increase after correcting the B_0-drift effects using oil tubes.
- The B_0 drift increases with time, which is more noticeable in long MRI acquisitions and should be always considered for MR thermometry.
- Study has limitations:
    - Temperature control can be done continuously, e.g., using fiber optics sensors.
    - GRE acquisition time can be faster, e.g., reducing the number of slices, planes, phase-encoding directions, among others.
- Improvements are already being made to the current experiments.

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Thank you!

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Phase unwrapping

Phase information:

- The MR signal recorded is complex and can be described as:

$$S(\vec{r},t) = \overbrace{A(\vec{r},t)}^{\text{Magnitude}}\, e^{\underbrace{i\varphi(\vec{r},t)}_{\text{Phase}}}$$

- The measured phase needs to be first rescaled (using the DICOM information) and present values that lie between $\pm\pi$ radians.

$$\boxed{\varphi(\vec{r},t) = \frac{\pi}{l_{pv}}(a_s\varphi_D(\vec{r},t) + a_i)}$$

- The measured phase is a wrapped representation of the true phase of the signal and needs to be unwrapped.

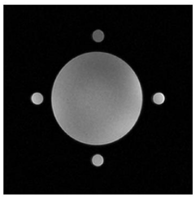
Phase unwrapping:

- The unwrapping operation aims to remove the 2π jumps from the wrapped phase and return the true phase φ_t of the signal:

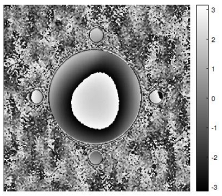
$$\boxed{\varphi_u(\vec{r},t) = \varphi_t(\vec{r},t) + 2\pi k(\vec{r},t)}$$

where $k(\vec{r},t)$ is the wrap count between $\varphi_u(\vec{r},t)$ and $\varphi_t(\vec{r},t)$, and is an integer number.

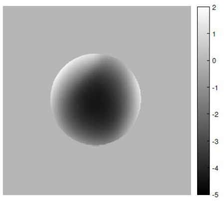
Magnitude



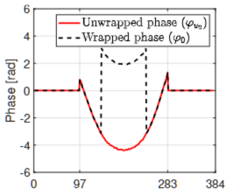
Rescaled phase (φ_r)



Unwrapped phase (φ_u)

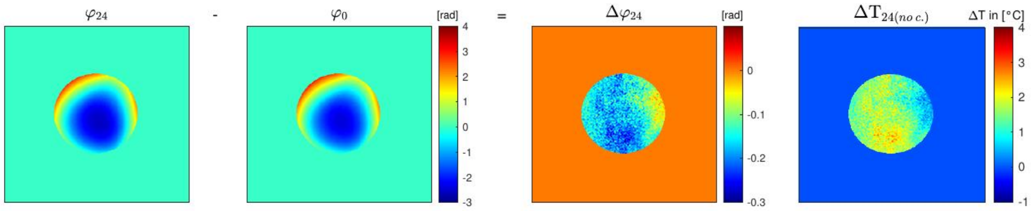


φ_r and φ_u



PRF shift method

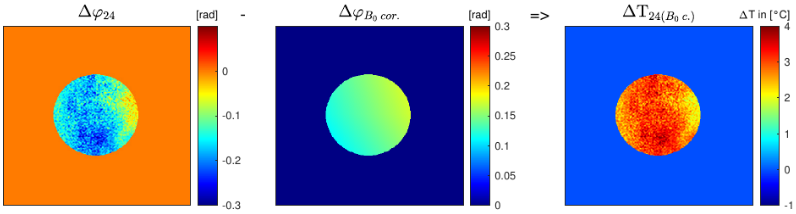
Temperature variation of the muscle-mimicking phantom:



$$\boxed{\Delta T_f = \frac{\varphi_f - \varphi_0}{\alpha\gamma B_0 TE}}$$

PRF shift improvements applying the drift corrections:

- Corrections are related to subject motion, dynamic temporal and spatial fluctuations of magnetic field due to heating of the magnet components during scanning, etc.



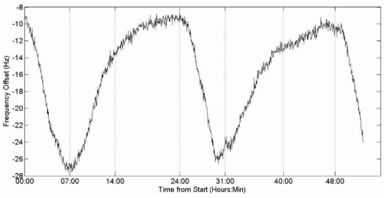
B_0 drift corrections:

$$\boxed{\Delta T_{f_c} = \frac{(\varphi_f - \varphi_0) - (\varphi_{oil_f} - \varphi_{oil_0})}{\alpha\gamma B_0 TE}}$$

PRF source of errors

1. **Field drift over time from the superconducting magnets**: For some 7T MR (Siemens) is in the range of 0.02 ppm/h. The drifts over time are the major difficulty for long experiments (1 to 2 hours, like some hyperthermia therapies).

-> **Influence:** This field drift alone account for a temperature error of ~2°C in a single hour.
-> **Correction:** One way to correct the error is to use external lipid reference probes such as oil. Lipid resonance frequencies show almost no temperature dependent shift. Thus, any change will be due to magnetic field drift.



Wyatt, et all. 2009.

Figure 4.
Long-term frequency drift of the 1.5T magnet, demonstrating the principally diurnal change as well as variable components that lead to poor predictability of the behavior.

2. **Temperature dependent change in electrical conductivity:**

-> **Influence:** This affects the PRFS thermal coefficient, and this effect is independent of TE.

-> **Correction:** Can be corrected with the double-echo PRFS-method, where two echo times are acquired and subtracted.

$$\Delta T_{corr} = \frac{(\Delta\phi_{\mathrm{H_20}}(TE_2) - \Delta\phi_{\mathrm{H_20}}(\mathrm{TE_1})) - (\Delta\phi_{oil}(TE_2) - \Delta\phi_{oil}(TE_1))}{\gamma\xi B_0(TE_2 - TE_1)}$$

PRF source of errors

3. **Spatial field drift:** causes shifts in reconstructed images.

-> **Influence:** If the amount of image shift is in the order of a voxel or greater, it will cause errors in the phase subtraction for PRFS.

-> **Correction:** 1. One could correct it on k-space data directly. Extensions to higher order polynomial corrections or more extensive prospective corrections may further improve accuracy in cases of more severe background drift. 2. Image shift artefacts over long scanning can be corrected using a rigid co-registration of incoming images to the baseline dynamic before subsequent polynomial fitting of the phase drift.

4. **Determination of the PRFS thermal coefficient (α):** Although some works show tissue-type independency for α value (ex. for ex vivo water-based animal tissues), there are some remaining discrepancies.

-> **Influence:** Most groups assume α between 0.0096-0.015 ppm/°C. For a given phase difference, this range of α may yield a 56% spread of values for T rise, (e. g. 6°C to 9.36°C).

-> **Correction:** One can characterise the material-specific α-parameter (depends on correct determination of probe location, low spatial field drift, etc). Once the temperature probe sensors are localised, the α-parameter can be extracted as the slope of the linear plot.

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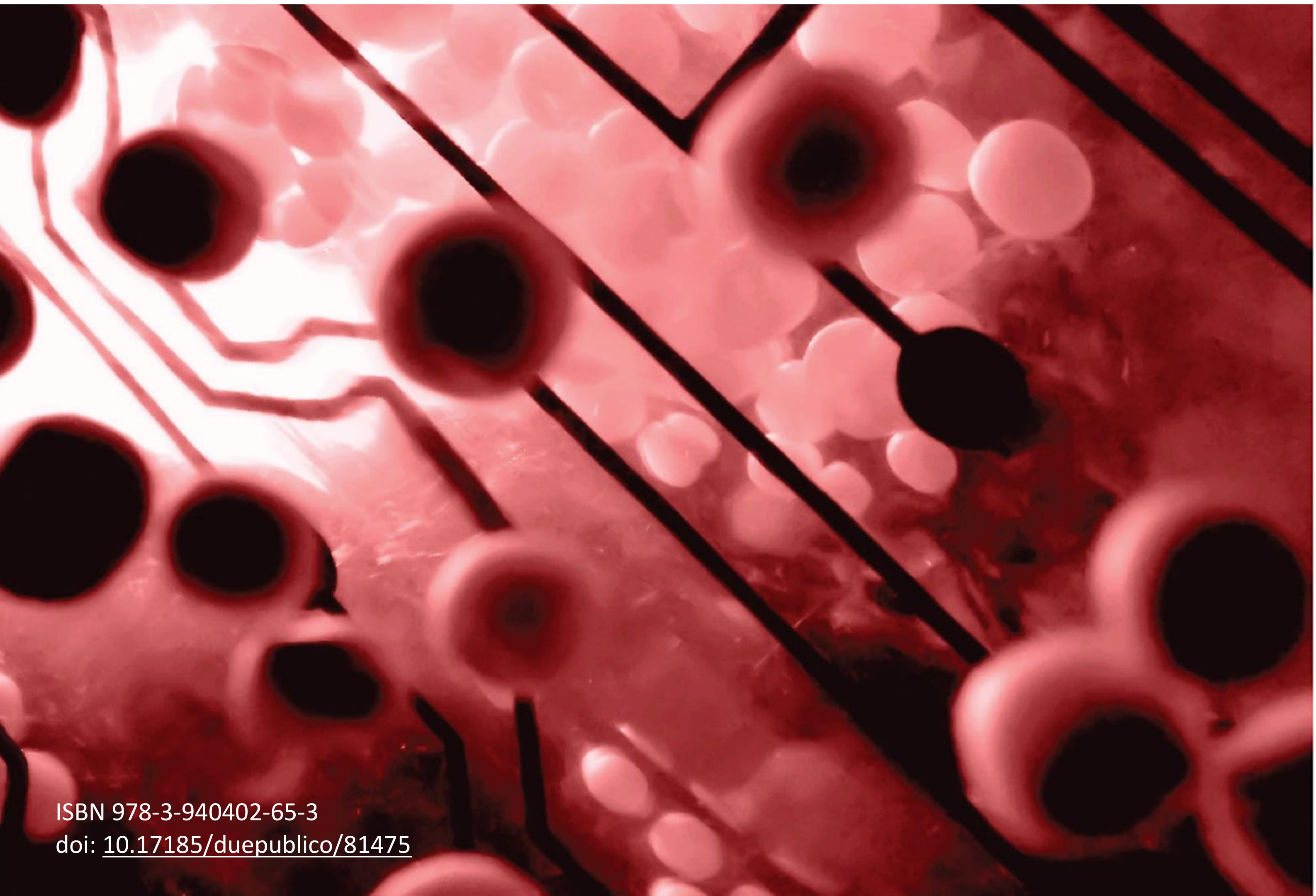

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