The compatibility of temporary pacemaker leads with magnetic resonance imaging – an ex vivo tissue study

R. Rzanny[1], A. Hansch[2], A. Pfeil[3], A. Gussew[1], S. Drobnik[4], and J. R. Reichenbach[1]

[1]Medical Physics Group, Department of Diagnostic and Interventional Radiology, Jena University Hospital, Jena, Germany, [2]Department of Diagnostic and Interventional Radiology, Jena University Hospital, Jena, Germany, [3]Department of Internal Medicine III, Jena University Hospital, Jena, Germany, [4]Department of Cardiothoracic Surgery, Jena University Hospital, Jena, Germany

Purpose

Temporary pacemaker leads have so far represented a contraindication to magnetic resonance imaging (MRI) [1]. However, MRI is an indispensable tool in the diagnosis and management of heart diseases as well as of cerebral injuries following cardiac surgery [2-3]. Hence, to date, MRI examinations have been performed only after removal of myocardial leads. However, at the same time, some studies suggest that implants are safe in MRI [4]. Risks associated with MR imaging in the presence of myocardial leads generally arise from the static main magnetic field, the rf electromagnetic field and the magnetic gradient fields [5-6]. One of the main risks arises from heating effects [7]. Additional risks are induced ventricular fibrillation [8], pacing distortions [7-9], or movement of the myocardial leads [10]. To estimate heating effects we investigated temperature changes in *ex vivo* tissue adjacent to implanted myocardial leads by ^{1}H-MRS. Temperature dependent shielding by hydrogen bonds cause a shift of the water proton resonance towards lower frequencies with increasing temperature [11]. The frequency separation of the water resonance to nearly temperature independent resonances, like total choline (tCho) and creatine/phosphocreatine (Cr/PCr) [12], was used to estimate the temperature.

Methods and Materials

Myocardial leads (Plastic tines model, Dr. Osypka GmbH, Rheinfelden, Germany) were implanted in 9 pig hearts. The hearts were subsequently investigated in a 1.5 T whole body scanner (Magnetom Sonata, Siemens Medical Solutions, Erlangen) using the standard receive head coil of the manufacturer. Following the first ^{1}H-MRS measurement to estimate the initial temperature, five different imaging sequences were applied: diffusion weighted EPI (SAR=0.2), T_1-SE (SAR=0.5 W/kg), T_2-TIRM (SAR=0.2 W/kg); gradient-echo EPI (SAR<0.1 W/kg) and ToF MR angiography (SAR=0.4 W/kg). The total acquisition time was approximately 16 min. Possible warming effects were estimated from a second MRS measurement. To maximize the absorbed rf energy a TSE sequence with a high SAR value (SAR=0.8 W/kg; TA=13 min) was applied next, followed by a third ^{1}H-MRS measurement for temperature estimation. One heart without implanted leads was investigated additionally by spectroscopy during several temperature cycles illustrated in Fig.1 by slow heating and cooling in a water bath between 20 and 40°C. The temperature in the cardiac tissue was controlled with a mercury-in-glass thermometer, positioned in the right ventricle. Single-voxel ^{1}H-MRS measurements (PRESS sequence, TR/TE=1500/135 ms; size: 40x10x23 mm; no water suppression) were started whenever the thermometer indicated temperatures of 23, 32 or 40°C and the bath temperature was adjusted to the heart temperature. Postprocessing of spectroscopic data included zero filling and phase correction as well as determination of peak frequencies and was performed with the software package MRUI (http://www.mrui.uab.es). Water frequencies were estimated directly from phase corrected spectra. Frequencies of tCho, Cr/PCr were estimated after water peak removal by HLSVD (Hankel Lanczos Squares Singular Values Decomposition) [13].

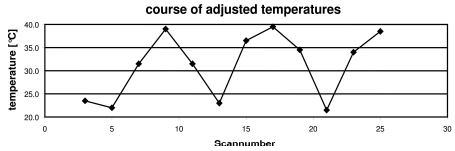

Fig. 1: *Course of adjusted temperatures to estimate thermometric shift coefficients*

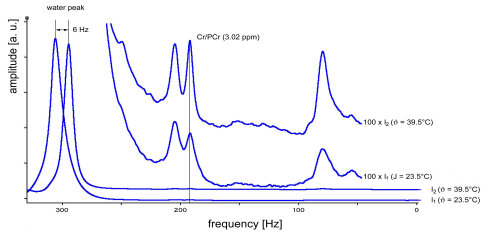

Fig. 2: *Illustration of the shift distance of the water peaks between 23.5 and 39.5 °C (0.095 ppm) adjusted by Cr/PCr at 3.02 ppm as temperature independent reference. Due to the excessive intensity of water the other resonances were presented with a magnification factor of 1500.*

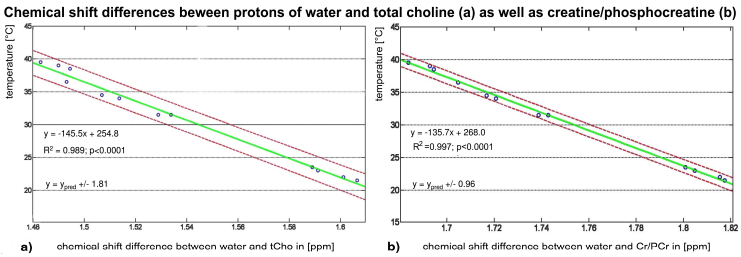

Fig. 3: *Correlation between temperature and chemical shift distance of the H_2O signal compared to tCho (a) and Cr/PCr (b) within the heating-cooling cycles illustrated by Fig. 1. Temperatures were measured with a mercury-in-glass thermometer. Chemical shift values were estimated by peak fitting using AMARES. The broken red lines indicate the confidence interval (95%).*

Results

As demonstrated in Fig.2, the signals of Cho (3.32 ppm), Cr/PCr (3.02 ppm) were easily identified in most spectra. As expected, at higher temperatures the water peak is shifted to lower frequencies. Fig.3 a, b shows the estimated shift difference between water and tCho (a) or Cr/PCr (b) with the temperature indicated by the mercury-in-glass thermometer during acquisition. Both diagrams show a linear correlation between shift and temperature values with a negative slope of -145.5 °C/ppm (tCho; p<0.0001; R^2=0.989) and -135.7 °C/ppm (Cr/PCr; p<0.0001; R^2=0.998) which correspond to temperature coefficients of the shift difference of -0.0069 and -0.0074 ppm/°C. The intervals of confidence (95%) indicate an accuracy of estimated temperatures of 1-2 °C based on the measured shift differences. All temperature values measured before and after the MR imaging session are summarized in Fig. 4. For three hearts the spectroscopic resolution of the last measurement was not sufficient to estimate temperatures (ϑ_3). Prior to MRI measurements hearts implanted with myocardial leads had a mean temperature of 21.08 ± 3.73°C. After application of standard clinical MRI sequences with a low SAR (SAR ≤ 0.6 W/kg), no significant temperature changes were observed. However, the majority of cases showed a small temperature decrease. After the high SAR sequence both small temperature increases and decreases were observed without a significant change.

Conclusions

Despite more difficult measuring conditions due to field distortions by the myocardial leads reliable temperature measurements by proton resonance frequency (PRF) thermometry seem to be feasible in *ex vivo* heart tissue. Temperature coefficients using tCho or Cr/PCr as a reference agree well with values of -0.007 ppm/°C presented by Mulkern et al. for rabbit [14] and -0.0067 ppm/°C by MacFall et al. for canine muscles [15]. The observed small temperature decrease after low SAR sequence application may be caused by evaporative cooling or adaptation to the room temperature. Although these investigations are only a preliminary step to evaluate the contraindication of myocardial leads, our results are encouraging to perform more detailed investigations, including size and geometry effects of the leads and to test the compatibility of leads by *in vivo* investigations in pig hearts.

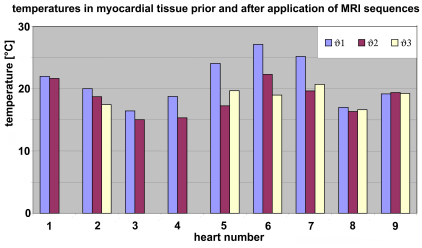

Fig. 4: *Estimated temperatures prior to MRI (ϑ_1), after application of the clinical standard sequences (ϑ_2), and after application of the high SAR sequence (ϑ_3) for all hearts.*

References

[1] Levine GN et al. Circulation 2007;116(24):2878-91.
[2] Laddis T et al. J Nucl Cardiol 2001;8(2):207-14.
[3] Gottesman RF et al. Semin Neurol 2008;28(5):703-15
[4] Shellock FG et al. Invest Radiol 2004;39(10):591-9.
[5] Shellock FG et al. Radiology 2004;232(3):635-52.
[6] Rezai AR et al. Invest Radiol 2004;39(5):300-3.
[7] Achenbach S et al. Am Heart J 1997;134(3):467-73.
[8] Peden CJ et al. Crit Care Med 1993;21(12):1923-8.
[9] Fontaine JM et al. Pacing Clin Electrophysiol 1998;21(6):1336-9
[10] Duru F et al. Eur Heart J 2001;22(2): 113-24.
[11] Ishihara Y et al. Magn Reson Med 1995;34(6):814-23.
[12] Young IR et al. J Magn Reson Imaging. 1998;8(5):1114-8.
[13] Pijnappel WWF et al. J Magn Reson 1992; 97 :122
[14] Mulkern RV et al. Med Phys. 1997 Dec;24(12):1899-906.
[15] MacFall JR et al. Med Phys 1996;23(10):1775-82.