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

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Abstract

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. 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 1H-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 nearby 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 1H-MRS measurement to estimate the initial temperature, five different imaging sequences were applied: diffusion weighted EPI (SAR=0.2), T2-SE (SAR=0.5 W/kg), T2-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 1H-MRS measurement for temperature estimation. One heart without implanted leads was investigated additionally by spectroscopy and post-processing of temperature dependent resonances. After application of the clinical standard sequence with a high SAR value (SAR=0.8 W/kg; TA=13 min) the heart was removed and temperatures were measured with a mercury-in-glass thermometer, positioned in the right ventricle. Single-voxel 1H-MRS measurements (PRESS sequence, TR/TE=1500/135 ms; slice: 40×10×23 mm; no water suppression) were started whenever the temperature of the heated heart was indicated by the mercury-in-glass thermometer during acquisition. Both diagrams show a linear correlation between shift and temperature indicated by the mercury-in-glass thermometer during acquisition. Both diagrams show a linear correlation between shift and temperature.

Results
As demonstrated in Fig.2, the signals of Cho (3.23 ppm) and Cr/PCr (3.02 ppm) were clearly 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 with a negative slope of -145.5 °C/ppm (tCho; p<0.0001; R²=0.899) and -135.7 °C/ppm (Cr/PCr; p<0.0001; R²=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. At temperatures measured before and after the MR imaging session were summarized in Fig. 4. For three hearts the spectroscopic resolution of the last measurement was not sufficient to estimate temperatures (�h). 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 (≤ 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 Mulker 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 feasibility of the method in in vivo investigations in pig hearts.

References