Oxygenation of mouse hearts in vivo at 17.6 Tesla

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Introduction

After myocardial infarction (MI) ischemic tissue passes through early apoptosis and necrosis, fibrosis and chronic inflammation. Finally, a thinned scar tissue can be observed. To monitor cardiac wound healing we used T_2^* imaging at high magnetic fields (17.6 T) to measure the oxygenation in the heart. T_2^* provides information about the oxygenation in the myocardial muscle and the oxygen tension provides functional information of the scar vascularization and viability. Oxygen is paramagnetic and leads to field inhomogeneities dependent on the amount of oxygen. From the difference of two T_2^* maps – one acquired while the animal breathes normal room air and the other while it breathes pure oxygen – we obtained a measure of the oxygenation in the tissue. Here we present the first results using mice in vivo.

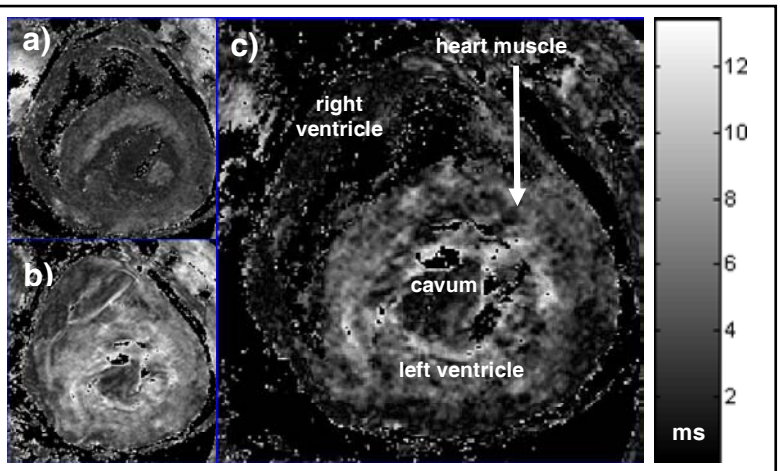

Fig. 1: a) T_2^*-map, mouse breathes normal air. b) T_2^*-map, mouse breathes pure oxygen. c) Oxygenation map (difference between a and b) of a healthy mouse heart. The in-plane resolution is 78 x 78 µm².

Methods

NMR imaging was performed on a Bruker Avance 750WB at 17.6 Tesla. To achieve a high spatial resolution we used a Micro gradient system with a gradient strength of 1 T/m. We worked with healthy mice and also with mice after a myocardial infarction. The myocardial infarcts were induced in female C57Bl/6 mice by ligation of the left coronary artery (LAD) and measurements were performed one week later. The animals were anesthetized with 1-1.5% isoflurane (air flow: 0.5 l/min). To eliminate motion due to breathing and heart beat we placed a pressure balloon on the chest of the mice and used the low frequency of the signal to skip motion phases of the gasping (approx. 1.2 – 3.3 Hz), and the high frequencies to trigger the MR sequence to the heart beat (approx. 8 Hz). The trigger delay (time between a trigger signal of the heart and the beginning of the diastole) was determined by using a CINE-sequence (in-plane resolution 156 x 156 µm², slice thickness 1 mm, 13 frames per heartbeat). From the beginning of the diastole we had 60 – 70 ms to obtain the data.
For T_2^* imaging a 2D gradient-recalled multi-echo sequence was used. We worked with different matrix sizes: 128 x 128 (FOV: 20 x 20 mm², spatial resolution 156 x 156 µm² in-plane), 256 x 256 (FOV: 20 x 20 mm², in-plane resolution 78 x 78 µm²) and 512 x 512 (FOV limited by the gradient hardware to 35 x 35 mm², in-plane resolution 68 x 68 µm²). The slice-thickness was 0.5 mm. After each RF excitation, 12 echoes were acquired and the T_2^* maps were obtained by mono-exponential fits. The available time during one diastole is long enough to acquire a complete k-space line for all 12 echoes, even for the 512 x 512 matrix size.

Results

T_2^*-values of the heart are very short at 17.6 Tesla (< 15 ms). In Figure 1a, a T_2^*-map of a healthy mouse heart breathing room air is shown. Figure 1b shows the same mouse breathing pure oxygen now. The difference map provides information about the oxygenation in the tissue (Figure 1c). The difference of T_2^* between normal air and pure oxygen is approx. 50%. In Figure 2, an oxygenation map of an infarcted mouse is shown. The thin scar in the anterior myocardial wall is clearly seen.

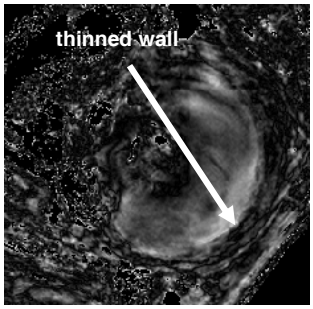

Fig. 2: Oxygenation map of a mouse heart with a myocardial infraction. The in-plane resolution is 78 x 78 µm².

Discussion & Conclusion

Oxygenation measurements are possible in vivo at high magnetic fields. The differences of the T_2^* values under different air conditions are significant. Spatial resolution has to be high enough to visualize the thinned myocardial scar, which is hard to see even at a resolution of 78 x 78 µm². Lower field strength may enable to improve SNR (e.g. 7 Tesla Biospec, gradient strength 870 mT/m) even if SNR is lost due to a lower B_0-field. Lower fields result in a longer T_2^* relaxation time which allows to acquire more gradient echoes.
Using high-resolution T_2^* imaging, the visualization of myocardial fibre structure is feasible resulting from susceptibility artefacts between myofibers and vessels (on a scale of 100µm) [1-4]. However, due to remaining motion artefacts caused by breathing and motion of the heart, structural differences are not yet visible (Figure 3). A minimal displacement between different echoes avoids the observation of the microstructure. Therefore, the next step will be to use artificial ventilation following intubation of the animals to eliminate respiratory motion.

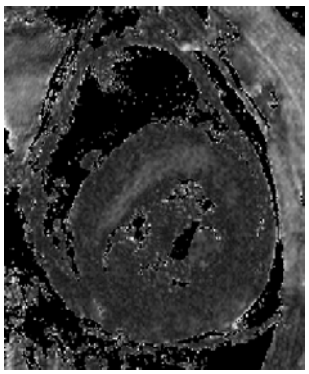

Fig. 3: T_2^*-map of the heart of a healthy mouse with the maximal in-plane resolution of 68 x 68 µm².

Acknowledgement

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References

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