First in vivo imaging of the mouse brain at 4.7 T using a subcentimeter HTS surface coil

Simon Auguste Lambert[1], Marie Poirier-Quinot[1], Ludovic De Rochefort[1], Jean-christophe Ginefri[1], and Luc Darrasse[1]
[1]*UMR 8081, IR4M, Univ Paris Sud, CNRS, Orsay, Paris, France*

Introduction

The use of a small surface coil is a way to increase the MR signal due to stronger magnetic coupling between the coil and the sample. Additionally, it filters out the RF noise from the rest of the body, leading to a large potential increase of the local signal-to-noise ratio (SNR). However the noise arising from the RF coil itself sets a limit to the gain available from the coil miniaturization [1]. Designing a 12-mm high-temperature superconducting (HTS) surface coil that features extremely low intrinsic noise have been shown to dramatically improve the SNR for *in vivo* mouse brain at 1.5 T, by more than an order of magnitude as compared to a room-temperature copper coil of similar geometry, which allowed to reach spatial resolutions of 60 μm^3 in a clinical MRI unit [2]. For field strengths of 4.7 T and above, cryocooled probes based on normal coil conductors at 30 K have been commercially developed and afford SNR gain factors of 2-4 [3]. Only few studies relate sensitivity improvement using superconducting coil at high fields and concern only non conductive and small samples, for which SNR gain are highly significant due to the absence of sample noise [4-8]. A particularly challenging issue with HTS coils for *in vivo* applications at high field is the design of sub-centimeter coils in order to achieve SNR gains overcoming the limits of normal conducting cryocooled probes. In this work the design and implementation of a HTS surface coil of 6 mm mean diameter, dedicated to in vivo imaging of the mouse at 4.7 T, is proposed, and the first *in vivo* mouse brain images are demonstrated.

Materiel and Method

According to [1] as B_0 increases sample losses increase more rapidly than coil losses. In order to benefit from the high sensitivity of HTS coil at 4.7 T the coil diameter has to be decreased less than 12 mm which is the optimized coil size used at 1.5 T. To determine the optimal size of the coil for *in vivo* application at 4.7 T we evaluated the different noise sources. The expected SNR gain is then calculated, for a HTS coil operated at 80 K, compared to a room temperature (RT) copper coil with the same geometry and using a semi infinite phantom with 0.66 $S.m^{-1}$ conductivity. This evaluation shows that the use of a 6 mm diameter HTS coil may yield the same SNR gain (of about 6) than the one obtained with a 12mm diameter HTS coil at 1.5 T. The SNR of an MR image can be expressed for a given RF coil geometry as a function of Q_l and Q_u, the loaded and unloaded quality factors of the RF coil, respectively, T_c, T_s, the RF coil and sample temperatures, respectively and F_{tot} the receiver channel noise factor [2,9] (eq. 1). Matching was adjusted to 50 Ω at an RF source level of –30 dBm, low enough to observe a linear response with the HTS coil. The quality factors Q_u and Q_l were extracted during this step using swept-frequency analysis of the RF coil impedance. The equivalent noise temperature of the receiver channel was evaluated using a spectrum analyser (Agilent- AS-N9000A). The identical HTS and copper surface coils were operated at 80 K and 298 K, respectively. Their design was a five-turn transmission line resonator with 6 mm diameter. Both HTS and copper coils were used in transmit/receive mode inductively coupled to the RF preamplifier. The HTS coil was mounted in a dedicated nitrogen cryostat [2]. Images were acquired on a 4.7 T scanner with a 3D gradient echo sequence, a matrix size of 256×256×100, a FOV of 15×15×12 mm, a bandwidth per pixel of 98Hz, TE/TR of 9/59 ms respectively and an acquisition time of 27 min. SNR estimations and measurements were performed on a non-conductive sample and a conductive sample with same load effect than mouse brain. Finally *in vivo* brain mouse imaging was acquired. A total of 4 nude mice of about 30 g were investigated under anesthesia induced by an intraperitoneal injection of 200 µL of diluted (1/5) pentobarbital (Sanofi Synthelabo Laboratory, Paris, France). Immediately after imaging, the mice were euthanized with an intraperitoneal injection of a pentobarbital over-dose.

$$SNR \propto \sqrt{\frac{Q_l}{\left(T_S(1-\frac{Q_l}{Q_u})+T_C\frac{Q_l}{Q_u}\right)\cdot F_{tot}}} \qquad (1)$$

Results

The measured SNR gain in imaging on the non-conductive, and the conductive phantoms using the HTS coil and the copper coil are of about 4.7± 0.97, and 3.8±0.92 in good agreement with the ones expected from eq. 2 and Q measurements (table 1) of 4.6, and 3.1 accounting for a F_{tot} of 12.6 (11 dB). Typical *in vivo* data acquired with the HTS coil are displayed in figure 1 for different anatomical mouse sites.

Sample	*Q* (Cu 298K)	*Q* (HTS 80K)	SNR_{gain}
Non conductive	129	2543	4,6
Conductive	121	1100	3,1
Mouse	126	1800	3,9

Table 1: *Quality factors of the HTS coil and the copper coil as a function of the load. The expected SNR gain of the HTS coil compared to the copper*

Conclusion, discussion and perspectives

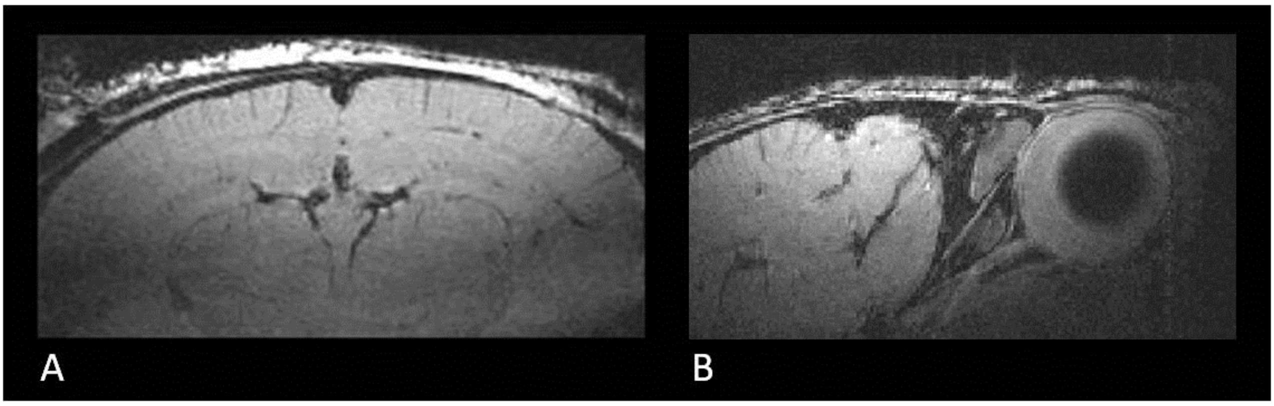

Figure 1 : *In vivo Axial (A) and Sagital (B) MR-images of the mouse brain made with the HTS coil.*

This is the first report of *in vivo* imaging using a subcentimeter cryocooled surface coil at 4.7 T. Complete RF characterization of the coil (Q_u, Q_l) and the receiver channel (F_{tot}) combined with a theoretical model [2, 9] allowed the validation of SNR gains. The SNR_{gain} for the mouse brain was about 4, so according to [10] the images shown in the present work are equivalent to those that could be obtained with a RT copper coil at 18 T. Morever the image SNR is mainly limited here by the poor noise figure of the receiver channel, due to a problem encountered with our T/R switch device in these first experiments. Assuming a better device with a total noise figure of 1 dB, the SNR gain is expected to be as high as 5.4 and the image SNR will improved by a factor of 10 compared to the preliminary images presented here. Finally, an additional gain of 1.5 in SNR might be obtained by decreasing temperature of the HTS coil by 14 K from 80 K as demonstrated in [11].

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Aknowledgement: We thank Bruker Biospin for granting a PhD application related to this work.