Radiofrequency (RF) coils fashioned from high-temperature superconductor (HTS) have the potential to increase the sensitivity of the magnetic resonance imaging (MRI) experiment by more than a dozen times compared to conventional copper coils. Progress, however, has been slow due to a series of technological hurdles. In this article, we present the developments that recently led to new perspectives for HTS coil in MRI, and challenges that still need to be solved. First, we recall the motivations for the implementations of HTS coils in MRI by presenting the limits of cooled copper coil technology, such as the anomalous skin effect limiting the decrease of the electric resistance of normal conductors at low temperature. Then, we address the progress made in the development of MRI compatible cryostats. New commercially available low-noise pulsed-tube cryocoolers and new materials removed the need for liquid nitrogen-based systems, allowing the design of cryogen-free and more user-friendly cryostats. Another recent advance was the understanding of how to mitigate the imaging artifacts induced by HTS diamagnetism through field cooling or temperature control of the HTS coil. Furthermore, artifacts can also originate from the RF field coupling between the transmission coil and the HTS reception coil. Here, we present the results of an experiment implementing a decoupling strategy exploiting nonlinearities in the electric response of HTS materials. Finally, we discuss the potential applications of HTS coils in bio-imaging and its prospects for further improvements. These include making the technology more user-friendly, implementing the HTS coils as coil arrays, and proposing solutions for the ongoing issue of decoupling. HTS coil still faces several challenges ahead, but the significant increase in sensitivity it offers lends it the prospect of being ultimately disruptive.

Keywords: high temperature superconductor, magnetic resonance imaging, MRI compatible cryostat, nonlinear properties of superconductors, HTS coil, cryogenics, imaging artefact
INTRODUCTION

The discovery of high-temperature superconductivity (HTS) by [1], paved the way for the application of superconductor-based technologies at temperatures above that of boiling nitrogen. Simultaneously, powerful, low vibration, reliable cryocoolers have been developed for applications at temperatures close to 90 K, the critical temperature of high-temperature superconducting YBa$_2$Cu$_3$O$_7$ (YBCO) which is the technologically most useful HTS material. In microwave technology, HTS has allowed the design of detectors and filters of unmatched sensitivity/selectivity and noise figures, suitable for satellite telecommunications [2]. A down-to-earth application of HTS materials is Magnetic Resonance Imaging (MRI). Here, the use of HTS materials can be envisioned both as magnet coils and as probes for the detection of the radiofrequency (RF) MR signal. The vanishing electric resistance of the superconducting MRI coil brings electric noise levels within reach that are unachievable with more traditional copper coils, even if the latter are cooled to much lower temperatures than the HTS coil; typically, the signal-to-noise ratio (SNR) can be increased more than a dozen times [3].

In spite of the promise held, the development of HTS radiofrequency MRI coils, starting from early prototypes in the 90s [4], has been slow. Interest in this technology faded in the 2000s due to technological deadlocks that seemed insurmountable at the time. These included drawbacks such as having to employ wet cryostats (e.g. using liquid nitrogen or LN$_2$), as well as the necessary decoupling of the RF detection coil (fashioned from a HTS material) from the (copper) RF transmission coil. Both issues induce important artifacts in MR images. Recently, however, a number of hurdles have been lifted, heralding new possibilities for HTS coils in MRI. In this paper, we review recent progress and challenges ahead.

First, we review the limitations of copper coils. Cryogenic RF coils become relevant when the dominant noise source in MRI is the RF coil used to collect the signal, i.e. in the case of nonconductive samples, or when conductive samples are probed with “small” surface antennas. With copper coils, noise reduction is readily limited by the intrinsic properties of the material. This limit can be significantly pushed by using superconductors, justifying the implementation for HTS coils in MRI with small samples or small RF coils.

We next present the progress made in the design of MRI compatible cryostats. Such cryogenic systems face the key challenges of ensuring full electromagnetic compatibility in DC and RF domains and user-friendly integration for routine practice in MR exams.

A recent study [5] has led to a much better understanding of how the static field ($B_0$)-inhomogeneities induced by magnetic flux expulsion in the HTS coil can be mitigated. In addition to their low intrinsic resistivity, HTS materials also present strong diamagnetism that can induce significant $B_0$-artifacts in MR images, such as localized signal loss and continuous phase shifts. Under the right circumstances, however, these can be avoided.

We then address the subject of the RF ($B_1$) coupling between the transmission coil (volume coil) and the HTS reception coil (surface coil). In the absence of magnetic coupling, the HTS coil acquires the MR signal with high sensitivity during reception and does not perturb the homogeneous RF excitation emitted by the transmission coil during transmission. This represents the targeted operational mode for HTS coils. In presence of coupling, however, the HTS coil picks up and locally amplifies the $B_1^+$-field generated by the transmission coil, possibly inducing important $B_1^+$-inhomogeneity artifacts in the MR images. These include large-scale signal amplitude modulations and localized π-phase shifts. To avoid the $B_1^+$-artifacts, a decoupling strategy must be implemented so as to deactivate the HTS coil during transmission. To illustrate this issue and its implications, this section will present the original results of an experiment implementing a decoupling strategy based on non-linearities in the RF response of the HTS material.

Finally, we discuss the perspectives for further improvements of HTS coil technology in view of novel MRI applications.

LIMITS OF COPPER COILS IN MAGNETIC RESONANCE IMAGING

Cooling the RF coil circuit at cryogenic temperatures to improve the sensitivity of nuclear magnetic resonance (NMR) detection has been an early idea [6]. From the fluctuation-dissipation theorem, the noise spectral density generated by a linear dissipative medium is proportional both to its temperature and to its power dissipation rate. A drastic reduction of the coil noise is thus expected as a result of drop of the coil resistance as the temperature is lowered. Conversely, neither the sample temperature nor its equivalent damping resistance into the coil are affected, so that the noise from the sample will eventually limit the improvement in sensitivity [7]. The earliest proofs of concept have been established with copper coils cooled to 4.2 K using liquid helium (LHe) and at 77 K with liquid nitrogen, respectively, to increase the sensitivity in $^{13}$C high-resolution spectroscopy [8], and in low-field clinical MRI [9]. A dramatic improvement in sensitivity was reported in NMR spectroscopy, whereas for MRI the gain was rather modest due to the large noise induced by the human body. A few years later, the potential of HTS ceramics for replacing copper had been demonstrated with prototypes of RF coils, at 77 K for low-field MRI [10], 10 K for MR microscopy at high field [4] and 25 K for $^1$H detection in a 400 MHz spectrometer [11]. Electrical losses of more than two orders of magnitude below that of copper were reported. Since these early achievements, however, the cooled-copper RF coil technology has spread widely in the markets of high-resolution NMR spectroscopy [12] and preclinical micro-MRI [13]. The replacement of copper by HTS materials, on the other hand, has remained a research topic in industrial and academic labs [3,14–18]. The objective of this section is to provide guidelines to assess the coil material performance in terms of ultimate SNR performance and report the limits achievable with copper.

Using the reciprocity principle [7], the contribution of the RF coil to the final signal-to-noise ratio in MR images can be expressed by the RF sensitivity factor $S_{RF}$:
The circuit is modeled by a self-inductance \( L \) and two sources in the coil and in the sample, respectively (with \( f_i \) the field amplitude in the coil during reception). Evaluating the interaction between the coil and a nuclear magnetic field only depends on the coil geometry. The measurement of the residual resistance ratio (RRR) [20]. The latter is due to the RF skin effect [\( \alpha \rho \rightarrow \infty \)]. The proximity factor \( \xi \) accounts for the proximity effect of the metal by conduction electrons in a normal conductor leading to the RF surface resistance \( \Omega \):

\[
\Omega_{sc} = \frac{\rho}{\delta} = \frac{\mu_0 \omega \rho}{2}
\]

with \( \delta \) the RF skin depth and \( \mu_0 = 4\pi \cdot 10^{-7} \) the magnetic permeability of the vacuum. Actually, at extremely low \( \rho \), \( \Omega_{sc} \) can be much larger than expected from Equation 4 because \( \delta \) decreases as \( \rho^{-1/2} \) while the electronic mean free path \( \lambda \) increases as \( \rho^{-1} \) (anomalous skin effect). Since a part of the accelerated electrons tends to leave the conducting skin layer, the effective surface resistance [21] increases significantly when

\[
\alpha = \frac{3}{2} \left( \frac{\lambda}{\delta} \right)^2 = \frac{3}{4} \mu_0 (\rho \lambda)^2 \omega \rho^{-3}
\]
becomes comparable or greater than 1, as showcased in Figure 1A (grey curves). The factor \( \rho(\omega) \) is an intrinsic constant for a conductor, close to \( 7.2 \cdot 10^{-16} \Omega \cdot \text{m}^2 \) for copper. When \( \omega \) increases well above \( 1/\Omega_{\text{sc}} \), it tends to a low asymptotic limit of about \( 3.4 \cdot 10^{-16} \omega^{-2/3} \). Thus, the anomalous skin effect only affects by a few percent the data at highest frequencies of Figure 1B, which relies on a RRR of 20, an order of magnitude for the RRR value of copper used in MRI probes. However, due to the fast decrease of \( \alpha \) with \( \rho \), it represents a severe limit for copper with larger RRR values.

Figure 1B also displays the normalized RF sensitivity factor expected for a typical HTS material at 77 K. A first key feature of HTS is the partition of the conduction electrons between a coherent Cooper pair state that carries the lossless supercurrent, and a dissipative contribution from excited quasiparticle states. A second is that the electromagnetic field is extinguished over a surface layer, the thickness of which corresponds to the frequency-independent London penetration depth \( \lambda_L \); in HTS, typically \( \lambda_L \approx 150 \text{ nm} \). The kinetic energy of the Cooper pairs is concentrated within this layer. Electrical losses are zero only for DC currents, when the Cooper pair supercurrent shorts the quasiparticle current. However, AC electric fields accelerate (and decelerate) both the Cooper pairs and the quasiparticles contained in the surface layer. Assuming that \( \mu_{\text{pl}} \omega \gg 1/\rho_{\text{qp}} \), with \( \rho_{\text{qp}} \) the quasiparticle resistivity, the theoretical surface resistance \( \Omega_{\text{sc}} \) is given by [2]:

\[
\Omega_{\text{sc}} \approx \frac{1}{2} \mu_{\text{pl}}^2 \omega^2 \rho_{\text{qp}}^{-2/3} \lambda_L^3. \tag{6}
\]

The depth \( \lambda_L \) increases rapidly as the temperature or magnetic field approach the critical values \( T_c \) or \( H_c \) above which the superconducting state disappears totally. Inversely, for “perfect” superconductors, the dissipative quasiparticle contribution is supposed to progressively vanish \( (\rho_{\text{qp}} \rightarrow 0) \) far below the critical limits. In HTS materials in particular, supplementary losses occur due to the motion of magnetic flux vortices, weak-link effects at grain boundaries, and the unconventional d-wave Cooper pairing mechanism, leading to complex and strongly anisotropic dependences of dissipation on the magnetic field and RF current [2]. In spite of fundamental limitations, much of the performance of a HTS material relies on its fabrication process though. For applications at 77 K, the best technology to date is that of thin-film YBCO epitaxially grown on single-crystal dielectric substrates such as lanthanum aluminate or sapphire, that exhibit extremely low tangent losses. Systematic measurements from 20 MHz to 20 GHz with temperatures between 30 and 77 K [22] show that the \( \omega^2 \) dependence expected from Equation 6 is only approached in the GHz range and is closer to a \( \omega^4 \) in the low-end frequency range up to a few hundred MHz, with a typical \( \Omega_{\text{sc}} \) of 0.5 \( \mu\Omega \) at 100 MHz. Wafers of up to 20 cm in diameter are currently available from manufacturers and can be accurately etched to elaborate monolithic resonators using the substrate as capacitance. Thin-film HTS coils with quality factors from a few 1,000 to several 100,000 have been reported for MR imaging applications [19].

Another consideration with cryogenic probes comes from the thermal insulation wall inserted between the sample and the low temperature RF coil. To preserve the field-of-view accessible with the coil, the thickness of the insulator needs to be accommodated with a larger coil radius. Consequently, looking back at Equation 1, this tends to mitigate the sensitivity gains predicted above by reducing the \( B_f/I \) factor. Indeed, from the Biot-Savart volume integral and considering coils with constant aspect ratios the \( B_f/I \) amplitude of surface coils is expected to decrease as \( 1/\alpha \) when the coil radius is adjusted to optimize the sensitivity at a target depth below the sample surface. The same \( 1/\alpha \) law also applies at the center of volume coils. Considering a volume copper coil as a typical example, the ultimate SNR gain at 30 K is about seven when compared to room temperature (Figure 1). For a nominal sample diameter of 5 mm and a realistic thickness of 2 mm for thermal insulation, a 30 K copper coil with a mean diameter of 9 mm would thus achieve a net gain of about four over a tightly adjusted RF coil at room temperature. Coping with even smaller samples would require an improved insulation to accommodate the larger surface-to-volume ratios, which intrinsically limits the advantages of cooled RF coils over room-temperature ones below a critical sample size. Since they offer a much larger ultimate SNR performance than copper, HTS materials can significantly alleviate this limit and bring substantial SNR improvements over conventional probes even with millimetric samples [17]. More generally, far more complex rules have to be considered with large conductive samples, with a general trend to limit the SNR gain reachable with cryocooling. In any case, however, HTS materials that provide much lower losses than their copper counterparts offer a promising perspective to design novel RF coil geometries such as actively screened or meta-structured configurations.

**PROGRESS IN MAGNETIC RESONANCE IMAGING COMPATIBLE CRYOSTATS**

The need for cryogenic means has long been seen as a serious bottleneck to the dissemination of cooled RF coil technology in MR applications. Many advances have been made during the last 20 years in the emerging field of superconductive electronics where the requirement for low-cost, compact, secure, and easy-to-use cryogenic devices has been a strong motivation [23]. However, MR detection meets severe additional constraints of compatibility with the presence of RF as well as of large static magnetic fields: no metallic parts can be present near the RF coil and no magnetic materials nor moving metallic parts can be in proximity to the magnet bore. Specific materials such as PVC, fiberglass composites, polystyrene or polyurethane foams, quartz, glass, or ceramics are generally involved in the cryostat body. Thus, initial developments [19] mostly relied on homemade cryogenic designs. First trials involved immersion of the coil circuit in LHe (4.2 K) or LN2 (77 K) [8,9], allowing intrinsically uniform and temperature-stabilized cooling of volume coils or large surface coils. To reduce mechanical and electrical instabilities due to boil-off and bubbling, the trend was then to replace the direct liquid immersion by a flow of cold...
nitrogen [24] or helium [25] gas, or by a cold finger made of high-performance dielectric material such as sapphire [11,26].

Another particular constraint that arises during the study of human subjects, animal models and biological samples is to maintain the living system at room temperature all the while keeping a small separation from the cooled RF coil. This can be provided either by a plain insulating wall [10], by a double-wall vacuum layer [25], or by a single-wall evacuated cryostat containing the RF coil [27]. The latter allowed the reduction of the insulator thickness to less than 1 mm, which is particularly suitable for high-resolution surface imaging such as that of the human skin and rodents in a preclinical setting. Nevertheless, using conventional superinsulation with metallized layers to block thermal radiation through the insulating window does not fit the requirement for RF transparency, so an amount of thermal leakage still occurs. Thus, insufflation of a warm stream of an inert gas such as N₂ into a reserved space along the outer cryostat surface has been applied to preserve the temperature stability of small living samples [24]. As pointed out in the previous section, the requirement for a minimized insulation thickness is not as severe with HTS coils as with copper coils since the formers offer much higher ultimate SNR gains.

More than 15 years after the early demonstrations of RF coils cooled by direct fluid immersion in the academic world, more elaborated closed-cycle cooled RF probes [28] have become available as standard accessories for analytical NMR spectroscopy [29]. In the new probes, typically, circulating helium gas transfers the cooling power from a distant two-stage Gifford-McMahon refrigerator to the RF coil, through heat exchangers at a regulated temperature ranging between 20 and 30 K. Temperatures of 20–30 K are low enough to improve the sensitivity of copper RF probes by a ratio of four when compared to room temperature, which enables 16 times faster acquisitions at constant SNR. Moreover, an integrated preamplifier cooled to below 80 K by the first stage of the Gifford-McMahon machine ensures a low noise contribution from the receiver, even with noiseless samples. Such new technology has been expanding rapidly since the late 1990s to face the high throughput issues for ensuring an efficient flow cooling.

The use of a refrigerant was eliminated in the cryogen-free system proposed by [40], later described by [42], and displayed in Figure 2. The cold source, provided by a commercial single-stage pulse tube cryocooler connected to a remote motor, is located directly on the non-magnetic cryostat vessel. The cold head is thermally coupled to the HTS coil by a series of high-conductivity pure aluminum flexible straps used in combination with copper and sapphire supports, as depicted in Figure 2B. This solution allows the transfer of sufficient cooling power to the RF coil; offers great adaptability to the RF coil bedding geometry; minimizes static and RF magnetic field perturbations induced by the system; and minimizes the transfer of mechanical vibrations to the HTS coil. Moreover, a non-magnetic heat exchanger tip anchored to the coil bed and a fully automated control system enable the control of the coil temperature to within 0.1 K, [41,42] Thus, the coil frequency can be tuned through temperature as a parameter. These aforementioned characteristics, as well as the reasonable weight (70 kg) and compact nature of the cryostat, allow an installation as close as possible to the measurement zone of the MRI scanner. In fact, this system is cryogenist-free—i.e. no cryogenic knowledge is required to use it -, autonomous, and adaptable to any MRI structure that stays in the examination room during the experiments.
MITIGATING THE EFFECTS OF HIGH-TEMPERATURE SUPERCONDUCTING COIL DIAMAGNETISM

Another recent advance was made by better understanding the interaction between the HTS coil and the static magnetic field $B_0$ used in MRI. In the presence of a magnetic field, superconductors spontaneously generate screening currents on their edges and surfaces to expel the magnetic flux. This induces static field ($B_0$)-inhomogeneities in the vicinity of the superconductor, which can result in $B_0$-artifacts in the MR images. In type-II superconductors, which include all HTS materials, magnetic flux can be readmitted in the form of quantized flux lines, or "vortices" [43]. Consequently, the magnetic properties of HTS materials are typically temperature dependent and hysteretic. Depending on the magnetic history of the material, notably whether it was cooled in the presence (field cooled) or the absence (zero-field cooled) of a magnetic field, it can display perfect or imperfect diamagnetism. As presented in the previous section, HTS coils for MRI applications have mostly been used at LN$_2$ temperature, i.e. sufficiently close to the superconducting transition temperature $T_c$ of the HTS material for its diamagnetism to be negligible.

However, with better-controlled experimental parameters and new cryostats achieving lower temperatures ($\sim 60$ K) this becomes an issue producing major $B_0$-artifacts on MR images in the vicinity of the HTS coil.

The static field inhomogeneity artifacts due to static field expulsion from HTS coil were recently investigated [5] for different cooling conditions and temperatures of the HTS coil. This was done by measuring the MR signal at 1.5 T of a water sample in the vicinity of a HTS coil: a 12 mm multi-turn Transmission Line Resonator [44] (MTLR) with a pair of six-turn YBCO spirals ($T_c = 86$ K) on both sides of a Al$_2$O$_3$ substrate. In the aforementioned study, the HTS coil was non-resonant to avoid RF $B_1^*$-artifacts while keeping intact the diamagnetic response of the superconductor. The coil was used in the cryostat described in Figure 2, and the normal to its surface was kept perpendicular to the static field $B_0$ during the whole experiment. A whole-body RF coil was used in transceiver mode. Standard 3D gradient-echo MR images were acquired after zero-field cooling (outside the MRI scanner), field cooling (inside the MRI scanner) and for different temperatures varying from 60 to 90 K.

In Figure 3, we present the main results of this study. Images acquired after zero-field cooling displayed important signal losses and phase perturbations (Figure 3A). The observed artifacts were predominant closer to the HTS coil and shared the same
characteristic dimensions. In comparison, images acquired after field-cooling the coil in the MR scanner were unaltered. The phase perturbation was also shown to have a strong temperature dependence (Figure 3B); the amplitude of the perturbation was continuously reduced as the temperature increased. Using a phenomenological model of the coil superconducting diamagnetism, it was shown that these artifacts were caused by a small yet unavoidable misalignment, less than 1°, of the HTS coil in the static field \( B_0 \), and that their temperature dependence mirrored the temperature dependence of the in-plane critical current of YBCO measured in comparable conditions [45].

Two important conclusions can be drawn from these results. 1) The \( B_0 \)-artifacts are linked to a low working temperature of the coil when compared to its \( T_c \), and 2) they can be mitigated at low temperature by field-cooling the HTS coil in \( B_0 \) at its working position in the MRI scanner.

**DECOUPLING THE HIGH-TEMPERATURE SUPERCONDUCTING COIL DURING TRANSMISSION**

Another possible source of artifacts comes from the magnetic coupling between the HTS surface coil and the volume coil used during transmission. Copper coil technology typically relies on diodes to deactivate the reception coil during transmission. Unfortunately, these standard techniques cannot be implemented with HTS materials, as the physical connections required for incorporating discrete components in the coil circuit degrade its quality factor and add too much noise. Decoupling is hence a major technological challenge for HTS coils. A solution can be found through the highly nonlinear electrical properties presented by HTS materials [46]. The quality factor \( Q \) of HTS coils depends on the power of the incident RF magnetic signal. Their \( Q \) of a few thousand during reception (low power stage), typically collapses to a few dozen during transmission (high power stage) [47,48]. These nonlinear properties, which can be tailored by fine engineering of the superconducting track composing the coil, can be exploited as a means to deactivate the HTS coil during transmission [49]. Here we present a study of these nonlinear properties. In a first step (Setup 1), we will present results of a RF characterization of a HTS coil realized in a low field environment (Earth magnetic field). These results will provide insight into the physical mechanism behind the
collapse of the coil quality factor $Q$. In a second step (Setup 2), we present this decoupling approach implemented in an MRI environment (Figure 2), allowing the material to pass from a zero-resistance state to a dissipative state during transmission [49]. In both setups, the experiments were conducted with a resonating ($\omega = 63.5$ MHz at 70 K) HTS surface coil of identical design as the one described in the previous section [5].

**Setup 1**—The HTS surface coil is used in a dedicated cryostat [50] shown schematically in Figure 4, cooled in Earth’s magnetic field and characterized with a VNA and a method inspired from [47]. The electrical parameters of the HTS coil are obtained through the measurement of its RF response using an inductive coupling approach involving a measurement probe overcoupled with the HTS coil. The measurement probe is a 10 mm diameter copper loop placed in front of the HTS coil at a 5 mm distance. The incident power produced by the VNA and transmitted to the HTS coil via the measurement probe is then converted in equivalent $B_1^*$ or $B_2^*$ values, depending on the RF amplitude with this experimental geometry. The HTS coil was left resonating at its own resonant frequency, different from the Larmor frequency, for the experiments involving Setup 1.

First, this setup simulates an MRI sequence. A high amplitude pulse with $B_1^* \in [0.02 \mu T, 0.4 \mu T]$ is applied with the measurement probe to mimic the RF pulse used during spin excitation (transmission). Then, a low amplitude pulse corresponding to $B_1^* = 0.02 \mu T$ is applied to reproduce the NMR signal during reception. The switch time between the two power levels is 12 μs. The HTS coil quality factor $Q$ was extracted as a function of $B_1^*$ (during the excitation pulse) for three temperatures $T = 60, 70,$ and 80 K.

Secondly, the setup performs a swept-frequency analysis where the modulus and phase of the compensated reflection coefficient $\rho_{\text{comp}}(\omega) = \rho(\omega) - \rho_0(\omega)$ are measured with the VNA as a function of the frequency $\omega$ for different incident $B_1^* \in [0.05 \mu T, 0.35 \mu T]$ and temperatures $T = 60, 70,$ and 80 K. Here, $\rho_0$ corresponds to a reference data set taken in the absence of the HTS coil, whereas $\rho$ integrates its contribution [47].

**Setup 2**—Then, the HTS coil is cooled inside the MR compatible cryostat [42] integrated into a 1.5 T clinical MRI (Achieva, Philips) shown in Figure 2. Fine frequency tuning of the HTS coil at $\omega_0$ is performed using a tuning coil [3], while impedance matching is obtained with a coupling coil (Figure 2D). The same coupling coil is then used to pick up the HTS coil signal during the experiment. The MRI sequence is a standard 3D gradient-echo using the parameters listed in Table 1. During transmission, a volume (whole-body) coil is used to generate rectangular RF pulses with nominal amplitude $B_1^* \in [0.08, 27 \mu T]$ and pulse durations $t \in [6.4 \mu s, 4.0 \mu s]$. The pulse duration $\tau$ is calculated consequently to keep the flip angle constant ($\alpha = \gamma B_1^* \tau$). Also, the imaging sample (Figure 5A) is filled with a 1 ml solution of water with a gadolinium-based contrast agent (Dotarem, $C = 2.5 \times 10^{-3}$ mmol/ml).

With this second setup, we first connected the coupling coil to the MRI acquisition RF channel to acquire the HTS coil signal during reception. A reference measurement was obtained using a room temperature copper coil of similar dimensions (Setup 2). Then, the imaging sample was removed and the coupling coil signal $V$ was connected to an oscilloscope to measure the response of the HTS coil during transmission as a function of $B_1^*$ at different temperatures $T \in [62 K, 80 K]$. A reference measurement was also acquired with the coupling coil only (i.e. without the HTS coil). The ratio $V/B_1^*$ was then used to track the evolution of the HTS coil quality factor $Q$. In addition to these measurements, the quality factor $Q$ of the HTS coil was also measured in this configuration using a VNA and an incident power producing a $B_1^* \sim 0.05 \mu T$.

Figure 6 presents the results obtained with Setup 1. Figure 6A displays the dependence between the $Q$ value of the HTS coil and $B_1^*$ for three working temperatures of 60, 70 and 80 K. We observe a transition from a superconducting state ($Q \approx 50000$) to a dissipative state ($Q \approx 100$) for $B_1^*$ values that are temperature dependent. At lower temperature (60 and 70 K), this transition occurs at $B_1^* \approx 0.2 \mu T$ whereas at 80 K (closer to $T_c$), $B_1^*$ is reduced by almost an order of magnitude to 0.04 μT. Also, we mention that the transition time (not shown here) between the two states was faster than $12 \mu s$.

The effects of the RF power on the HTS material are shown in Figures 6B–G, which displays the modulus and phase of $\rho_{\text{comp}}$, noted respectively $|\rho|$ and $\phi$ thereafter, for different incident $B_1^*$ and working temperatures. First, at low $B_1^*$, $|\rho|$ and $\phi$ display the expected behavior for classical RF coils, except for the temperature dependence of the resonance frequency; at 60 and 70 K, the coil resonates close to 63.5 MHz, which decreases to 63.1 MHz at 80 K. This frequency shift, which is due to the temperature-induced increase of the kinetic inductance of the superconductor, is predominant when $T_c = 86$ K is approached. Secondly, at higher $B_1^*$, we observe a collapse of $|\rho|$ and a deformation of $\phi$ that are also temperature dependent. At 80 K, $|\rho|$ and $\phi$ simply flatten as $B_1^*$ increases. This is to be compared with the behavior observed at 60 and 70 K, where $|\rho|$ becomes strongly deformed at $B_1^* = 0.35 \mu T$ and $\phi$ showcases a discontinuity (inset of Figures 6C,E).

In Figure 5, we present the results acquired with Setup 2. The spatial inhomogeneities observed in the MR images of Figures 5B–C are due to a non-uniform flip angle of the spins over the
sample. This is produced during transmission by the re-emission of the spatially non-uniform $B_1^+$ field generated by the surface coil, the amplitude of which is proportional to its quality factor $Q$. In the absence of artifacts, the MR signal would appear relatively uniform over the sample. For a copper coil presenting linear electrical properties (Figure 5C), the magnitude of the artifacts does not depend on the value of $B_1$, as expected at a given flip angle $\alpha$. For the HTS coil, however, the artifacts in Figure 5B are more important at lower $B_1^+$. This indicates that increasing $B_1^+$ increases the resistance of the coil, and consequently reduces its quality factor $Q$ during transmission. This behaviour is well observed when we measure the signal of the coupling coil.
 durante transmission. As shown in Figure 5D, the ratio $V/B_1^2$, which is proportional to $Q$, decreases as a function of $B_1^2$ of almost one order of magnitude between 0.05 and 10 μT at 62 K. In comparison, this coefficient is almost constant for all $B_1^2$ values without the HTS coil (ref. curve in black on Figure 5D). For values of $B_1^2$ below 1 μT, $Q$ displays a strong temperature dependence, which is not the case anymore for higher $B_1^2$. This dependence was also observed with the $Q$ measurement of the HTS coil inside the magnet performed with the VNA, typically with $Q \sim 3000$ at 60 K and $Q \sim 2560$ at 70 K.

These two setups studied the nonlinear electrical properties of the HTS coil in Earth’s magnetic field (Setup 1) and in a field of 1.5 T (Setup 2). We observed differences between the behavior of $Q$ in Earth’s field and at 1.5 T, in agreement with a previous study (Lambert RSI 2008). This difference can be explained by vortex dynamics, which we assume to be the principal dissipation mechanism in our experiment [51]. The AC current circulating in the HTS coil, depending on $(B_1^2)^2$, is responsible for energy dissipation in the HTS material and leads to the increase of its electrical resistivity (i.e. its surface impedance). In Earth’s field, vortex penetration is weak and does not, unless the temperature closely approaches $T_c$, affect the HTS material response much. At low $B_1^2$, the vortices remain pinned and the RF current is not strong enough to induce large vortex displacements. Consequently, the surface impedance of the coil remains constant, leading to $Q \approx 50000$ for all working temperatures, in good agreement with precedent measurements [Lambert RSI 2008 [3]]. At higher $B_1^2$, the effect of hysteretic vortex motion in the HTS material becomes significant, inducing an increase of the electrical resistance, and a decrease of the $Q$ values that amounts to one order of magnitude between 60 and 70 K. This is even more pronounced with an increase of temperature; we observe a decrease of two orders of magnitude of $Q$. At 1.5 T, the critical current density is lower, and the vortices are more easily set in motion by RF currents even at low $B_1^2$. This phenomenon becomes more pronounced as the temperature of the HTS material is increased. Consequently, $Q$ is halved between 60 and 80 K. In contrast, for $B_1^2 \geq 1\mu$T, $Q$ does not depend on the temperature anymore. This is most probably due to a complete penetration of the material by the vortex lines carrying $B_1^2$; once the RF field completely penetrates the material, there is no further change in field contrast associated with $B_1^2$, so that all related artifacts are now independent of temperature. In the experimental results, we observed that the value of $B_1^2$ at which the superconducting material transits from the superconducting state to the dissipative state depends on temperature, but also on the value of the static field. This can be explained by the temperature, field, and field-orientation dependence of the critical current density of the HTS material, as studied by [45], in comparable conditions.

**DISCUSSION**

In this work, we reviewed some of the challenges and recent advances made in the implementation of HTS coil in MRI. The key feature of HTS coils is that they do not share the lower limit on resistivity experienced by copper coils, although the $\omega^2$ proportionality law between their electrical noise and working frequency limits their potential for applications in high-field MRI. HTS coils do, however, share the same extrinsic limitations as cryogenic copper coils. To take full advantage of their benefits, the necessity for the sample to be non-conductive, or at least not to overload the coil (c.f. Equation 2), restrains the imaging volume to small dimensions for *in vivo* applications at standard medical imaging fields (i.e. up to a few milliliters at 1.5 and 3 T), as discussed in the literature [19]. As for cryogenic copper coils, this limits the prospects of implementations of HTS coils as volume coils. For larger samples, they could nevertheless be implemented as coil arrays, which would allow HTS coils to image far greater fields-of-view while keeping advantageous SNRs. The issue of decoupling the array elements from each other entails substantial work, but a solution could be based on the work of [52], or that of [14]. Considering the small dimensions of HTS coils, they could only be used for surface imaging in clinical MRI for skin [53] or extremities imaging, for instance. In preclinical MRI, they will probably come as a natural successor to cryogenic copper coils, as they are well suited for weakly conductive small biological samples such as mice and rats, or *ex-vivo* biological samples [54]. The high sensitivity of HTS coils also opens up the way to high-resolution MRI (<100 μm) at relatively low fields (<3 T) when compared to the more usual high fields used in preclinical MRI (4.7 T and higher). Since the MR relaxation times depend on $B_0$, there is a strong motivation to investigate animal models directly in clinical magnetic fields, since this approach greatly facilitates the transfer of MR protocols from preclinical to clinical imaging. Hence, this limits the need for additional investments.

In order for these applications to be developed, the HTS coil must be integrated into a cryostat that achieves both MRI compatibility and good thermal insulation between the cryogenic probe and the room temperature imaging subject. Although this has proven to be an important technological challenge, recent years have seen the design of new and more robust cryostats, with the notable development of the cryogen-free and autonomous system that we described [42]. While we focused our attention primarily on closed-cycle technology, onboard open-cycle cooling approaches dedicated to small RF coils show promising perspectives to address the challenge of cooling high-density flexible HTS coil arrays, i.e. arrays where the relative positions of the coil elements can be modified. Such individual onboard cooling units or, alternatively, distributed refrigerant streams, could be implemented respectively with a compact N₂ Joule-Thomson refrigerator [55] or a microfluidic LN₂ exchanger [56]. The actual challenges of such a device are improving the compactness and reducing the cooling time while keeping a standalone, user-friendly, and cryogen-free solution. Compactness will be achieved by reducing the dimensions, and therefore the weight of the device. Consequently, this will reduce the heat stored in the system at room temperature as well as the thermal resistance between the cooling system and the coil, allowing for faster cooling or less cooling power. This cannot be accomplished without a fine optimization of the thermal capacity of the thermal link with the cooling power of the cryocooler at all stages of cooling. These improvements will
bring the device even closer to the MRI measurement zone and impose having even more non-magnetic material entering its composition. Using new polymeric materials—robust, lighter, and compatible with vacuum and MRI scanners—, will fulfill both requirements.

The question of the optimal working temperature for HTS coils remains open. On the one hand, we have seen that working at temperatures significantly lower than \( T_c \) may be associated with circumstantial static field inhomogeneity artifacts caused by the superconducting diamagnetism of the coil; these could be avoided by field cooling the coil in the MRI magnet at its working position. On the other hand, working at lower temperatures should lead to a better sensitivity factor \( (S_{RF}) \) of the HTS coil. Therefore, regarding superconducting diamagnetism, working at a lower temperature can be envisioned if these artifacts are consistently avoided. For most applications, this implies frequent temperature cycling of the coil in-between experiments or after a change in geometry. This requires a practical implementation involving quick, localized and uniform heating of the HTS coil inside the cryostat. Such a heating system could work by radiative, resistive, or conductive heating but without any interference with the MRI experiment, and would solve an important issue in the implementation of HTS coils by providing the experimental conditions required to obtain better images with this technology.

The issue of tuning and decoupling of the HTS coil also comes into play for the selection of its working temperature. Until now, our studies were conducted using HTS coils working at around 80 K in transceiver mode. As stated earlier, the HTS coil design is chosen so as to resonate close to the Larmor frequency, and a fine tuning of its resonance frequency is achieved by positioning a closed-circuit copper loop \[3\]. Thanks to the new cryostat, this fine tuning can now be performed by a fine control of the temperature, as proposed by \[42\], since the coil resonance frequency depends on the temperature \[39\]. Also, matching the HTS coil to the source impedance was achieved by a contactless matching technique using an inductive pick-up loop tuned at the Larmor frequency. The coupling coefficient is manually adjusted, modifying the distance between the pick-up loop and the HTS coil, to reach the optimal 50 \( \Omega \) presented at the input preamplifier. This matching step can be made more user-friendly by the integration of an automated system based on piezoelectric motors inside the cryostat \[57\].

In parallel, until the issue of the magnetic coupling between the HTS coil and the transmission volume coil is solved, the former can still be used in transceiver mode (i.e. as the transmitter and receiver) Typically, the transmit level of the coupled volume coil is manually adjusted to account for the presence of the HTS coil, which effectively becomes the source of the RF excitation flipping the sample magnetization during transmission. From the perspective of the volume coil, this leads to flip angle values of one to two orders of magnitude smaller than the ones currently used in MR protocols in order to achieve a real \( \pi/2 \)-flip angle in the region of interest (ROI) These precautions are necessary to limit the \( B_1^+ \)-artifacts caused by the HTS coil non-uniform magnetic field. Although this corresponds to a suboptimal implementation of HTS coil, high quality MR images can still be obtained in this operational mode \[3\]. However, the presented investigations were all based on gradient-echo protocols, and imaging a larger area was realized at the cost of a nonuniformly weighted T1 contrast. Many of the sequences requiring a uniform flip angle to assess a quantitative T1 or T2 contrast, for instance, are not usable in this condition. Consequently, achieving decoupling with HTS coil will open up a new world of applications for this technology.

To that end, the approach illustrated in our experimental results exploits the non-linear electrical response of HTS materials to minimize the quality factor during transmission. Although the coil under study did not reach resistance levels sufficient to achieve proper decoupling, it did provide a proof of concept for this technique. To go further, the HTS coil have to reach its dissipative state for lower RF \( B_1^+ \) field values which implies finding ways for the coil current to reach the critical current density of the HTS material sooner. This must be accomplished without degrading the coil sensitivity during reception. This objective can be achieved either by reducing the critical current density notably through temperature, as seen in Figure 5 and providing another argument for working at higher temperature, or by increasing the current density in the coil by a fine engineering of the superconducting track \[49\]. This can be performed by adding constrictions of various geometries in the coil design and modifying the HTS material thickness.

**CONCLUSION**

HTS coil for MRI has come a long way in the last 30 years. Cryogenics have advanced sufficiently to allow the design of MRI compatible cryostats that allow one to reliably operate HTS coils. In addition, we now better understand how HTS coils interact with the superposed magnetic fields \( B_0 \) and \( B_1 \) used in MRI, allowing for a better integration of the superconducting coil in the MR environment. This technology still faces several challenges before it is ready for practical applications in bio-imaging. However, the significant increase in sensitivity it offers lends it the prospect of being ultimately disruptive.

**DATA AVAILABILITY STATEMENT**

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

**AUTHOR CONTRIBUTIONS**

AL: coordinated the writing of the paper; was involved in the redaction of many sections of the article; and acquired parts of the data shown. GA and BB: designed the MRI compatible cryostat shown in the paper; and both were involved in the writing of the section “Progress in MRI compatible cryostats” which displays their work. JB and CvdB: helped design the experiment—Setup 2—shown in the paper; were involved in the writing of the section “Decoupling the HTS coil during transmission;” provided expertise during the discussion of the experimental results LD:
wrote the state-of-the-art sections for “Limits of copper coils in MRI” and “Progress in MRI compatible cryostats” MPQ: coordinated the research project; involved in the design of the experiments shown in this work; main writer of the section “Decoupling the HTS coil during transmission.” All authors actively contributed to the writing of this paper.

FUNDING

Experiments were performed on the 1.5 T MRI platform of CEA/SHFJ, affiliated to the France Life Imaging network and partially funded by the network (Grant ANR-11-INBS-0006), under Grant No. ANR-14-CE17-0003 (SupraSense project) and LabeX LaSIPS (ANR-10-LABX-0040-LaSIPS).

ACKNOWLEDGMENTS

The authors thank Dr. Jean-Christophe Ginefri for his contribution to the development of HTS coils within our research group. The authors also thank Rose-Marie Dubuisson for her help with the 1.5 T MRI scanner platform, and Georges Willoquet for his work on the electronics implemented in both cryostats used in this work, as well as Isabelle Saniour and Michel Geaehl for their past contributions to this project.

REFERENCES

1. Bednorz JG, and Müller KA (1986) Possible high-Tc Superconductivity in the Ba-La-Cu-O System. Z für B - Condens Matter 64, 189–93. doi:10.1007/BF01303701
2. Lancaster MJ (1997, Passive Microwave Device Applications of High-Temperature Superconductors). Passive Microwave Device Applications of High-Temperature Superconductors. United Kingdom: Cambridge University Press doi:10.1017/CBO9780511526688
3. Poirier-Quinot M, Ginefri J-C, Girard O, Robert P, and Darrasse L (2008) Performance of a Miniature High-Temperature Superconducting (HTS) Surface Coil for In Vivo Microimaging of the Mouse in a Standard 1.5T Clinical Whole-Body Scanner. Magn Reson Med 60, 917–27. doi:10.1002/mrm.21605
4. Black RD, Roemer PB, Mogro-Campero A, Turner LG, and Rohling KW (1993) Decoupling the HTS coil during transmission.
5. Black R, Early T, Roemer P, Mueller O, Mogro-Campero A, Turner L, et al. (2015) Advancing Cardiovascular, Neurovascular and Renal Magnetic Resonance Imaging in Small Rodents Using Cryogenic Radiofrequency Coil Technology. Front Pharmacol 7, 255. doi:10.3389/fphar.2015.00255

14. Wosik J, Xue L, Xie L-M, Kamel MR, Nesteruk K, and Bankson JA (2007) Superconducting Array for High-Field Magnetic Resonance Imaging. Appl Phys Lett 91, 183503. doi:10.1063/1.2801384
15. Nouls JC, Ixenzen MG, Gureley HP, and Johnson GA (2008) Design of a Superconducting Volume Coil for Magnetic Resonance Microscopy of the Mouse Brain. J Magn Reson 191, 231–8. doi:10.1016/j.jmr.2007.12.018
16. Lin I-T, Yang H-C, and Chen J-H (2012) Enlargement of the Field of View and Maintenance of a High Signal-To-Noise Ratio Using a Two-Element High-T C Superconducting Array in a 3T MRI. PLoS One 7, e42509. doi:10.1371/journal.pone.0042509
17. Ramaswamy V, Hooker JW, Withers RS, Nast RE, Brey WW, and Edison AS (2013) Development of a 13C-Optimized 1.5-mm High Temperature Superconducting NMR Probe. J Magn Reson 235, 58–65. doi:10.1016/j.jmr.2013.07.012
18. Laistler E, Poirier-Quinot M, Lambert SA, Dubuisson R-M, Girard OM, Moser E, et al. (2015) In Vivo MR Imaging of the Human Skin at Subnanoliter Resolution Using a Superconducting Surface Coil at 1.5 Tesla. J Magn Reson Imaging 41, 496–504. doi:10.1002/jmri.24549
19. Darrasse L, and Ginefri JC (2003) Perspectives with Cryogenic RF Probes in Biomedical MRI. Biochimie 85, 915–37. doi:10.1016/S0300-9084(03)00091-6
20. Matula RA (1979) Electrical Resistivity of Copper, Gold, Palladium, and Silver. J Phys Chem Reference Data, 8(4), 1147–298. doi:10.1063/1.555614
21. Sondheimer EH (1952) The Mean Free Path of Electrons in Metals. Bull Phys Soc Japan 5, 230–73. doi:10.1126/science.8430331
22. Wang Y, Su HT, Huang F, and Lancaster MJ (2007) Measurement of YBCO Thin Film Surface Resistance Using Coplanar Line Resistor Techniques from 20 MHz to 20 GHz. IEEE Trans Appl Supercond 17, 3632–9.
23. Anders S, Blamire MG, Buchholz F-I, Crété D-G, Cristiano R, Febvre P, et al. (2015) European Roadmap on Superconductive Electronics - Status and Perspectives. Physica C: Superconductivity 470, 2079–126. doi:10.1016/j.physc.2010.07.005
24. McFarland EW, and Mortara A (1992) Three-dimensional NMR Microscopy: Improving SNR with Temperature and Microcoils. Magn Reson Imaging 10, 279–88. doi:10.1016/0730-725X(92)90487-K
25. Black R, Early T, Roemer P, Mueller O, Mogro-Campero A, Turner L, et al. (1993) A High-Temperature Superconducting Receiver for Nuclear Magnetic Resonance Microscopy. Science. 259, 793–5. doi:10.1126/science.8430331
26. Miller JR, Zhang K, Ma QY, Mun IK, Jung KJ, Katz J, et al. (1996) Superconducting Receiver Coils for Sodium Magnetic Resonance Imaging. IEEE Trans Biomed Eng 43, 1197–9. doi:10.1109/30.544344
30. Ratering D, Baltes C, Nordmeyer-Massner J, Marek D, and Rudin M (2008) Performance of a 200-MHz Cryogenic RF Probe Designed for MRI and MRS of the Murine Brain. Magn Reson Med 59, 1440–7. doi:10.1002/mrm.21629
31. Baltes C, Radzwill N, Bosshard S, Marek D, and Rudin M (2009) Micro MRI of the Mouse Brain Using a Novel 400 MHz Cryogenic Quadrature RF Probe. NMR Biomed. 22, 834–42. doi:10.1002/nbm.1396
32. Junge S (2012) Cryogenic and Superconducting Coils for MRI. eMagRes 1, 505–14.
33. Ginefri J-C, Poirier-Quinot M, Girard O, and Darrasse L (2007) Technical Aspects: Development, Manufacture and Installation of a Cryo-Cooled HTS Coil System for High-Resolution In-Vivo Imaging of the Mouse at 1.5 T. Methods 43, 54–67. doi:10.1016/j.ymeth.2007.03.011
34. Hu B, Varma G, Randell C, Keesee SF, Schaeffter T, and Glover P (2012) A Novel Receive-Only Liquid Nitrogen (sHILoLN) [218] -Cooled RF Coil for High-Resolution In Vivo Imaging on a 3-Tesla Whole-Body Scanner. IEEE Trans Instrum Meas 61, 129–39. doi:10.1109/TIM.2011.2157575
35. Lin I-T, Yang H-C, and Chen J-H (2013) A Temperature-Stable Cryo-System for High-Temperature Superconducting MR In-Vivo Imaging. Plas One 8, e61958. doi:10.1371/journal.pone.0061958
36. Sánchez-Heredia JD, Baron R, Hansen ESS, Laustsen C, Zhurbenko V, and Ardenkjaer-Larsen JH (2020) Autonomous Cryogenic RF Receive Coil for 13 C Imaging of Rodents at 3 T. Magn Reson Med 84, 497–508. doi:10.1002/mrm.282113
37. Vester M, Steinmeyer F, Roas B, Thummes G, and Klundt K (1997) High Temperature Superconducting Surface Coils with Liquid Nitrogen or Pulse Tube Refrigeration (Absstr.). in Vancouver, BC: ISMRM 5th Scientific Meeting and Exhibition, doi:10.4095/219009
38. Darrasse L, Chiron L, Ginefri J-C, Poirier-Quinot M, Trollier T, Ravex A, et al. (2004) Cryosonde IRM : antenne IRM supraconduite pour la microscopie des régions superficielles du corps humain et des petits modèles animaux. ITBM-RBM 25, 254–9. doi:10.1016/j.itbm.2004.09.003
39. Lambert S, Ginefri J-C, Poirier-Quinot M, and Darrasse L (2013) High-temperature Superconducting Radiofrequency Probe for Magnetic Resonance Imaging Applications Operated below Ambient Pressure in a Simple Liquid-Nitrogen Cryostat. Rev Scientific Instr 84, 054701. doi:10.1063/1.4820947
40. Authelet G, Poirier-Quinot M, Ginefri J-C, Bonelli A, and Baudouy B (2017) Conceptual Design of a Cryogen-free μMRI Device. in JOP Conf Ser Mater Sci Eng, 12122.
41. Authelet G, Bonelli A, Poirier-Quinot M, Ginefri JC, Sanioiu I, and Baudouy B (2019) All Polymer Cryogen Free Cryostat for μ-MRI Application at Clinical Field. in JOP Conf Ser Mater Sci Eng, 12156.
42. Sanioiu I, Authelet G, Baudouy B, Dubuisson R-M, Jourdain L, Willoquet G, et al. (2020) A Temperature-Controlled Cryogen Free Cryostat Integrated with Transceiver-Mode Superconducting Coil for High-Resolution Magnetic Resonance Imaging. Rev Scientific Instr 91, 055106. doi:10.1063/1.5143107
43. Tinkham M (1996) Introduction to Superconductivity. 2nd ed. eds J Shira and E Castellano New-York: McGraw-Hill.
44. Serfaty S, Haizza N, Darrasse L, and Kan S (1997) Multi-turn Split-Conductor Transmission-Line Resonators. Magn Reson Med 38, 687–9. doi:10.1002/mrm.1910380424
45. Diez-Sierra J, López-Dominguez P, Rückerta H, Rûkel M, Hänisch J, Khan MZ, et al. (2020) High Critical Current Density and Enhanced Pinning in Superconducting Films of YBa2Cu3O7–δ Nanocomposites with Embedded BaZrO3, BaHfO3, BaTiO3, and SrZrO3 Nanocrystals. ACS Appl Nano Mater 3, 5542–53. doi:10.1021/acsanm.0c00814
46. Shen ZY (1994) High-temperature Superconducting Microwave Circuits. Boston, MA: Artech House Publishers.
47. Girard O, Ginefri J-C, Poirier-Quinot M, and Darrasse L (2007) Method for Nonlinear Characterization of Radio Frequency Coils Made of High Temperature Superconducting Material in View of Magnetic Resonance Imaging Applications. Rev Scientific Instr 78, 124703. doi:10.1063/1.2825241
48. Geethal M (2018) Découplage de détecteurs radiofréquences superconducteurs à très haute sensibilité pour la micro-imagerie par résonance magnétique. https://tel.archives-ouvertes.fr/tel-01880203/document (Accessed March 20, 2021).
49. Sanioiu I, Authelet G, Baudouy B, Dubuisson RM, van der Beek CJ, Darrasse L, et al. (2019) Novel Passive Decoupling Approach for High Resolution HTS RF Coils Based on the Nonlinear Electrical Properties of Superconductors. Rotterdam: ESMRMB 2019. in 36th Annual Scientific Meeting. Springer.
50. Sanioiu I, Geethal M, Briatico J, Beek CJvan der, Willoquet G, Jourdain L, et al. (2020) Versatile Cryogen-free Cryostat for the Electromagnetic Characterization of Superconducting Radiofrequency Coils. EPJ Tech Instrum 7, 1–11. doi:10.1140/epjti/s40485-020-00055-2
51. Clem JR, and Sanchez A (1994) Hysteric AC Losses and Susceptibility of Thin Superconducting Disks. Phys Rev B 50, 9355, 62. doi:10.1103/physrevb.50.9355
52. Frass-Kriegl R, Laistler E, Hosseinezhad S, Schmid AI, Moser E, Poirier-Quinot M, et al. (2016) Multi-turn Multi-gap Transmission Line Resonators - Concept, Design and First Implementation at 4.7 T and 7 T. J Magn Reson, 273, 65–72. doi:10.1016/j.jmr.2016.10.008
53. Ginefri J-C, Durand E, and Darrasse L (1999) Quick Measurement of Nuclear Magnetic Resonance Coil Sensitivity with a Single-Loop Probe. Rev Scientific Instr 70, 4730–1. doi:10.1063/1.1150142
54. Poirier-Quinot M, Frasca G, Wilhelm C, Luciani N, Ginefri J-C, Darrasse L, et al. (2010) High-resolution 1.5-Tesla Magnetic Resonance Imaging for Tissue-Engineered Constructs: A Noninvasive Tool to Assess Three-Dimensional Scaffold Architecture and Cell Seeding. Tissue Eng C: Methods 16, 185–200. doi:10.1089/tenc.2009.0015
55. Wright AC, Song HK, Elliott DM, and Wehrli FW (2005) Use of a Joule-Thomson Micro-refrigerator to Cool a Radio-Frequency Coil for Magnetic Resonance Microscopy. Rev Scientific Instr 76, 014301. doi:10.1063/1.1823440
56. Koo C, Godley RF, Park J, McDougall MP, Wright SM, and Han A (2011) A Magnetic Resonance (MR) Microscopy System Using a Microfluidically Cryo-Cooled Planar Coil. Lab Chip 11, 2197–203. doi:10.1039/c1lc20056a
57. Li Z, Willoquet G, Guillot GV, Hosseinnejhad S, Jourdain L, Poirier-Quinot M, et al. (2016) Study of Two Contact-Less Tuning Principles for Small Monolithic Radiofrequency MRI Coils and Development of an Automated System Based on Piezoelectric Motor. Sensors Actuators A: Phys. 241, 176–89. doi:10.1016/j.sna.2016.02.008

Conflict of Interest: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Copyright © 2021 Labbé, Authelet, Baudouy, van der Beek, Briatico, Darrasse and Poirier-Quinot. This is an open-access article distributed under the terms of the Creative Commons Attribution License (CC BY). The use, distribution or reproduction in other forums is permitted, provided the original author(s) and the copyright owner(s) are credited and that the original publication in this journal is cited, in accordance with accepted academic practice. No use, distribution or reproduction is permitted which does not comply with these terms.