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Tunable Ultrahigh Dielectric Constant (tuHDC) Ceramic Technique to Largely Improve RF Coil Efficiency and MR Imaging Performance

Wei Chen,

Center for Magnetic Resonance Research (CMRR), Department of Radiology, University of Minnesota, 2021 6th St. SE., Minneapolis, MN 55455, USA

Byeong-Yeul Lee,

Center for Magnetic Resonance Research (CMRR), Department of Radiology, University of Minnesota, 2021 6th St. SE., Minneapolis, MN 55455, USA

Xiao-Hong Zhu,

Center for Magnetic Resonance Research (CMRR), Department of Radiology, University of Minnesota, 2021 6th St. SE., Minneapolis, MN 55455, USA

Hannes M. Wiesner,

Center for Magnetic Resonance Research (CMRR), Department of Radiology, University of Minnesota, 2021 6th St. SE., Minneapolis, MN 55455, USA

Maryam Sarkarat,

Pennsylvania State College of Engineering, University Park, PA 16802, USA

Navid P. Gandji,

Department of Neurosurgery, Pennsylvania State College of Medicine, Hershey, PA, 17033, USA

Sebastian Rupprecht,

Department of Neurosurgery, Pennsylvania State College of Medicine, Hershey, PA, 17033, USA

Qing X. Yang,

Department of Neurosurgery, Pennsylvania State College of Medicine, Hershey, PA, 17033, USA

Michael T. Lanagan

Pennsylvania State College of Engineering, University Park, PA 16802, USA

Abstract

This work introduces an innovative magnetic resonance (MR) imaging technology to incorporate radiofrequency (RF) coil(s) with permittivity-tunable ultrahigh dielectric constant (tuHDC) ceramics by controlling the ceramic temperature to optimize and significantly improve RF coil transmission and reception efficiencies, MR imaging sensitivity and signal-to-noise ratio (SNR). The tuHDC ceramics made of composite barium strontium titanate (BST) compounds ($\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$) have low dielectric loss and very high permittivity tunability from 2,000 to 15000 by varying the ceramic temperature between 0°C to 40°C to achieve an optimal permittivity for

MR imaging application. We demonstrated for the first time the proof of concept using the BST-based tuHDC-RF-coil technology for improving MR spectroscopic imaging performance of ^{17}O nuclide at 10.5 Tesla (T) at a low ceramic temperature and ^{23}Na nuclide at 7T at room temperature. We discovered a large and spatially independent noise reduction under an optimal ceramic temperature, which synergistically resulted in an unprecedented SNR improvement. Large improvements were also demonstrated for ^1H MRI on a 1.5T clinical scanner using the same ceramics. The tuHDC-RF-coil technology is highly robust, flexible and cost-effective; it presents a technical breakthrough to significantly improve imaging sensitivity and resolution for broad MR imaging applications; and is critical to advance biomedical and neuroscience research, and improve diagnostic imaging.

Keywords

magnetic resonance (MR) imaging (MRI); MR spectroscopy imaging (MRSI); ultrahigh permittivity tunable ceramics; imaging sensitivity; imaging denoising; ultrahigh field (UHF)

I. INTRODUCTION

Magnetic resonance (MR) imaging (MRI) provides an important means for biomedical research and clinical diagnosis. The traditional MRI measures the resonance signal of the water protons for imaging tissue structures and functions. On the other hand, *in vivo* MR spectroscopy (MRS) or imaging (MRSI) methods detect ^1H or X-nuclear (e.g., ^{17}O , ^2H , ^{13}C , ^{23}Na and ^{31}P) resonance signals for assessing biochemicals and metabolites in tissues of interest. However, MR imaging faces a fundamental challenge: low intrinsic signal-to-noise ratio (SNR) owing to an extremely small energy difference between the excited and ground states of the nuclear spins. The intrinsic SNR of MR signal detected by a radiofrequency (RF) coil from a sample can be expressed as [1, 2]:

$$\text{SNR} = \frac{\text{MR signal from sample}}{\text{Noise}} \propto \gamma^2 \cdot B_0^\beta [\sin(\alpha)] \frac{B_1^-}{\sqrt{P_s + P_c}}; \alpha = \gamma \cdot \tau \cdot B_1^+ \cdot V, \quad (1)$$

where γ is the gyromagnetic ratio, B_0 is the static magnetic field strength, α is the RF pulse flip angle produced by the RF coil transmission magnetic field B_1^+ with a RF pulse duration τ and voltage V ; B_1^- is the RF coil reception magnetic field. The noise is determined by the dissipated RF power in the sample, P_s , and in the receive coil, P_c ; which is proportional to the electric conductivity (σ) and electric field (E) produced by the RF coil in the sample [3].

The ultrahigh field (UHF) strength for advancing the MR imaging and RF coil technology [4–7] has now reached 10.5 Tesla (T) for a whole-body human scanner [8]. The UHF technology is an effective but expensive approach to improve SNR according to the relationship of $\text{SNR} \propto B_0^\beta$, where β ranges approximately from 1 for ^1H MRI/MRS(I) to 7/4 for low- γ ^{17}O and ^2H MRS(I) [9–13]. Alternatively, it is critical to exploit new RF engineering solutions for further increasing MR imaging sensitivity and spatiotemporal resolution at a given field strength. The approach to integrate high or ultrahigh dielectric

constant (HDC or uHDC) materials or ceramic block(s) with a RF coil could effectively improve $\mathbf{B}_1$ fields and SNR for MR imaging application [14–23].

In this work, we introduce an innovative RF engineering approach that varies the material temperature to largely tune the permittivity of uHDC (tuHDC) ceramics made of composite barium strontium titanate (BST) compounds; and demonstrate for the first time by controlling the ceramic temperature (within $\pm 15^\circ\text{C}$ of room temperature: $T_{\text{Room}} = 20^\circ\text{C}$ for this study), we were able to adjust the tuHDC ceramic permittivity for a very wide range from ~ 2000 to >10000 for optimizing the RF coil performance and achieving unprecedented improvements for ^1H and X-nuclear MRI/MRSI applications on 1.5T-10.5T human scanners.

The basic principle of the tuHDC-RF-coil technology follows Maxwell's equations to describe the electric-magnetic fields generated by a RF coil [14–16, 20, 24, 25],

$$\nabla \times \mathbf{B}_1 = \mu\sigma\mathbf{E} + i\mu\epsilon_r\epsilon_0\omega\mathbf{E} \quad \text{Ampere's Law} \quad (2)$$

$$\nabla \times \mathbf{E} = -i\omega\mathbf{B}_1 \quad \text{Faraday's Law} \quad (3)$$

where $\mathbf{B}_1$ is the RF magnetic field that determines $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$, μ is the magnetic permeability, ϵ_r is the relative permittivity or dielectric constant, ϵ_0 is the permittivity of free space ($=1$), i is the unit complex number, ω is the RF operation or Larmor resonant frequency. The first term ($\sigma\mathbf{E}$) on the right side of Eq. 2 is the conductive current density ($\mathbf{J}_c$), and the second term ($\epsilon_r\epsilon_0\omega\mathbf{E}$) is the displacement current density ($\mathbf{J}_d$); they are the sources attributing to $\mathbf{B}_1$ fields. The displacement current is commonly disregarded since $\mathbf{J}_d$ is very small in air with $\epsilon_r \approx 1$ at the RF regime. Since $\mathbf{J}_d$ is proportional to ϵ_r , uHDC material with, e.g., >1000 -fold ϵ_r of air would largely enhance the $\mathbf{J}_d$ contribution and increase $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$, which leads to large improvements of SNR ($\propto \mathbf{B}_1^+$) and RF transmission efficiency ($\propto \mathbf{B}_1^+$).

II. METHOD

Temperature-dependent permittivity and dielectric loss of the BST-based tuHDC ceramics

The relative permittivity (ϵ_r) of the tuHDC ceramics in the paraelectric state is temperature dependent according to the Curie-Weiss Law:

$$\epsilon_r \approx \frac{C}{T_b - T_c} \quad (4)$$

where C is the Curie constant, T_c is the Curie temperature that can be found in the literature for various permittivity material and composition [26], and T_b is the tuHDC ceramic block temperature ($^\circ\text{C}$). C was approximately 90000 for the BST-based tuHDC ceramics used in this study based on the parameters of $T_c \approx 0^\circ\text{C}$ and $\epsilon_r \approx 4500$ measured at the room temperature of 20°C .

To conduct a proof of concept study, we utilized the unique property of the composite barium strontium titanate (BST) materials ($\text{Ba}_x\text{Sr}_{1-x}\text{TiO}_3$, x : barium to strontium ratio) with low dielectric loss for constructing the tuHDC ceramics in which the ϵ_r value can be varied by changing the ceramic temperature (T_b). The BST ceramic with $x=0.6$ has a very high permittivity and tunability (ϵ_r from ~ 2000 to >10000), and a low dielectric loss tangent ($\tan\delta < 0.03$) at the RF operation frequencies for MRI applications above T_c , in which the ceramics are in the paraelectric state [26–28].

The dielectric loss and its dependence on T_b and MR imaging operation frequency (ω) are important characteristics of the tuHDC ceramics, which need to be considered in design and optimization for MR imaging applications. In the paraelectric state and the ceramic temperature range of practical importance, the dielectric loss tangent obeys:

$$\tan\delta = \frac{\epsilon''}{\epsilon_r} \propto \omega T_b^m \epsilon_r^{1.5} \quad (5)$$

where ϵ'' is the imaginary part of the dielectric permittivity and m is a power constant [26].

The dielectric properties of each BST ceramic blocks were characterized by the resonant post method [29]. Permittivity was determined from the resonant frequency of the TE₀₁₁ resonant mode (100–200 MHz range) and the sample geometry. The dielectric loss was determined from the Q-factor of the resonant mode.

The uHDC and tuHDC ceramics used in this study

One rectangular uHDC ceramic block (10.2 cm $\times$ 7.7 cm, 1.4 cm thickness) was made of ferroelectric material of lead zirconate titanate (PZT: $\text{Pb}(\text{Zr}, \text{Ti})\text{O}_3$) (TRS, State College, PA, USA), which has an ultrahigh dielectric constant value ($\epsilon_r \approx 1000$). It had a relatively high $\tan\delta$ value in the RF operation frequency range of 120–125 MHz [30], and used for testing ^{31}P MRS at 7T ($\omega=120$ MHz). Another curved uHDC ceramic block (13 cm width, 21.4 cm arc length, 2.3 cm thickness) was made of PZT ($\text{Pb}(\text{Zr}_{0.5}\text{Ti}_{0.5})\text{TiO}_3$) ferroelectric material with a low $\tan\delta$ (≈ 0.05) and very high permittivity ($\epsilon_r \approx 1200$) (HyQ Research Solutions, LLC, State College, PA). This curved uHDC block was used to test ^{31}P MRSI at 7T for phantom and human brain studies. The permittivity values of these two uHDC ceramic blocks are fixed and not temperature sensitive, thus, non-tunable.

Development of functional tuHDC ceramics with very high permittivity-tunability involves many manufacturing parameters, which include mixing powders of desired material in right proportions (formulation), compressing them into desired shapes and then heat treat in high temperature (sintering) for an optimal period of few days through a three-step (formulation-sintering-characterization) iterative process [31–33]. The ceramics made by the composite BST materials can reach an extremely high dielectric constant at room temperature, which depends on the barium to strontium ratio (x , $0 < x < 1$) in the formulation of $\text{Ba}_x\text{Sr}_{1-x}\text{TiO}_3$ [34]. The adjustment of variable x can control the ceramic permittivity and dielectric loss, and could shift the material phase from the paraelectric to ferroelectric state for $x > 0.7$ at the room temperature. Ferroelectric materials will experience a significant dielectric relaxation near or above 100 MHz, and degrade MR imaging performance. In this study, we fabricated

the BST-based tuHDC ceramics ($\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$, $x=0.6$) for the applications of ^{17}O MRSI at 10.5T, ^{23}Na MRSI at 7T and ^1H MRI at 1.5T operated at the paraelectric state and a relatively low resonance frequency of <100 MHz, in which ultrahigh permittivity values are required for achieving best performance.

Specifically, two circular tuHDC ceramic blocks (8 cm diameter, 2.1 cm thickness) were made of $\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$ with a very low $\tan\delta$ (< 0.03 for 10–30°C), which depends on ω , T_b and ϵ_r according to Eq. 5. The permittivity values of the two ceramics were $\epsilon_r \approx 4500$ and 4700, respectively, at room temperature (20 °C). Within the temperature range of interest: i.e., above the Curie temperature ($T_c \approx 0$ °C) and below 40 °C in this study, the tuHDC ceramic is in the paraelectric state with $\epsilon_r > 2000$ at $T_b \approx 40$ °C and $\epsilon_r > 15000$ near the Curie temperature.

^{31}P MRS imaging studies of phantom and human brain at 7T

All ^{31}P MRSI experiments were carried out on a Siemens 7T whole-body/90-cm bore human scanner (MAGNETOM, Erlangen, German). A ^{31}P - ^1H dual-nuclei transverse electromagnetic (TEM) volume coil (inner diameter = 28 cm, coil length = 10 cm) was used, which consists of eight RF coil rungs at the ^1H operation frequency (297 MHz) and other eight elements at the ^{31}P operation frequency (120 MHz) driven by the quadrature RF power mode [35, 36]. The ^{31}P and ^1H coil channels were tuned and matched in the presence or absence of the uHDC ceramic block. The ^{31}P MRSI data were acquired using the three-dimensional (3D) chemical shift imaging (CSI) based on the Fourier Series Window (FSW) technique [35, 37]. The experimental protocol consisted of paired acquisitions in the presence and absence of the uHDC ceramic, respectively, and care was taken to place the imaging object in the same position within the TEM volume coil for the two comparative acquisitions with proper coil tuning and matching.

A spherical phantom (15-cm diameter) was used for 7T ^{31}P MRSI tests, which was filled with inorganic phosphate (Pi: 50 mM), sodium chloride (50 mM), and gadolinium contrast agent (0.05 mM) for shortening the longitudinal relaxation time (T_1) of Pi to ~300 ms. The curved low-loss uHDC block was positioned at the bottom of the phantom inside the ^{31}P - ^1H TEM head coil as shown in Fig. S1b. Multiple 3D ^{31}P -CSI data sets were acquired under the fully relaxed condition with the following parameters: repetition time (TR) = 2s, RF (hard) pulse width = 1 ms, 3D phase-encoding matrix = $9 \times 9 \times 7$, spectral bandwidth = 5 kHz, field of view (FOV) = $40 \times 40 \times 36$ cm³, number of complex data points of FID = 1024, varied RF pulse flip angles by adjusting RF pulse voltage, and number of signal average = 1.

In vivo ^{31}P MRSI experiments were performed on three healthy human volunteers, and all procedures were in accordance with the ethical standards of the institutional and national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards. The protocol was approved by the Institutional Review Board (IRB) of the University of Minnesota, and informed consents were obtained from all individual participants in this study.

The *in vivo* 3D ^{31}P CSI data were acquired under partially saturated condition with variable RF pulse voltages including an Ernst flip angle ($\alpha = 42^\circ$) for the phosphocreatine (PCr)

resonance ($TR = 1.0$ s, 3D matrix size = $9 \times 9 \times 7$, spectral bandwidth = 5 kHz, FOV = $20 \times 20 \times 18$ cm³, number of complex data points = 800, RF hard pulse width = 1 ms, two signal averages). The nominal voxel size of 3D ³¹P CSI was 12.7 ml, and total acquisition time was 10.7 minutes for collecting each CSI data set.

Another ³¹P MRS test was conducted using a small (1.7-cm diameter) spherical Pi phantom and a ³¹P surface RF coil (15.6-cm diameter) in the presence or absence of the rectangular-shape uHDC block made of PZT material with a high dielectric loss. The ³¹P MR spectra were obtained with varied RF pulse voltage.

Post-processing of the ³¹P MR spectra included the temporal zero-filling of FIDs to 2048 data points without using the line broadening. Spectrum analysis was performed using a Lorentzian-curve fitting with AMARES fitting algorithms [38] and the spectral noise level was calculated using a standard deviation (SD) of spectral baseline, i.e., the chemical shift regions without the presence of the ³¹P resonance signal.

The 10.5T ¹⁷O MRSI and 7T ²³Na MRSI phantom studies

All ¹⁷O MRSI experiments were carried out on a Siemens 10.5T whole-body/88-cm bore human scanner (MAGNETOM, Erlangen, Germany) with a single-loop ¹⁷O RF surface coil (15-cm diameter) operated at 60.6 MHz. A cylindric-shape bottle (9 cm inner diameter and 7.2 cm height) filled with saline water (77 mM NaCl) was used. For the tuHDC studies, the circular tuHDC ceramic block was placed at the RF coil center underneath the water phantom and above the temperature controller, and it was surrounded by Styrofoam for heating insulation (see the experimental setup shown in Fig. S1a). The RF coil plane was positioned at the similar level (see the dashed line in Fig. S1a) as the bottom surface of the tuHDC block. The 3D ¹⁷O-CSI data sets were acquired under fully relaxed condition with the following parameters: $TR = 0.2$ s, RF (hard) pulse width = 2 ms, 3D matrix size = $9 \times 9 \times 7$, spectral bandwidth = 30 kHz, FOV = $10 \times 10 \times 10$ cm³, number of complex data points of FID = 1024, various RF pulse flip angles by adjusting RF pulse voltage, and no signal averaging.

The similar MRSI sequence, parameters and phantom setup as employed in the ¹⁷O MRSI study at 10.5T were adapted for conducting the ²³Na MRSI phantom study on a Siemens 7T whole-body/90-cm bore human scanner (MAGNETOM, Erlangen, Germany). The ¹⁷O RF coil employed for the 10.5T study was re-tuned/re-matched to the ²³Na resonance frequency of 78.6 MHz operated at 7T.

¹H MRI phantom study on a clinical 1.5T MRI scanner

All ¹H MRI experiments were carried out on a Siemens 1.5T whole-body/70-cm bore human scanner (Magnetom Aera, Erlangen, Germany) using a commercial ¹H human head coil with 16-channel receiver array operating at 64 MHz. Due to the clinical MRI setup, the ¹H coil was not re-tuned/matched with/without the tuHDC ceramic. A similar cylindrical saline phantom and the same tuHDC ceramic as employed for the 10.5T ¹⁷O and 7T ²³Na MRSI studies were used. The ceramics were refrigerated to ~ 10 °C before taking out for the experiment during which its temperature slowly increased towards equilibrium. The ¹H MRI data were acquired using 2D FLASH (Fast Low Flip Angle Shot) sequence with

following parameters: TR/TE = 40 ms / 2.5 ms; FA = 20 degree; FOV = 256 mm × 256 mm; Slice thickness = 10 mm; image matrix = 128×128×7; number of averages = 2; multiple images were acquired with varied RF pulse voltages for $\mathbf{B}_1$ mapping.

$\mathbf{B}_1$ mapping

The MR imaging signals were used to calculate $\mathbf{B}_1$ images. The relative magnitude of $\mathbf{B}_1^+$ is inversely proportional to the RF pulse transmitter voltage for reaching a 90° RF pulse flip angle ($V_{90^\circ} \propto 1/\mathbf{B}_1^+$), and the relative magnitude of $\mathbf{B}_1^-$ is proportional to the maximum MR imaging signal reached at 90° RF pulse flip angle under the fully relaxed condition ($S_{90^\circ} \propto \mathbf{B}_1^-$). These relationships were employed to generate representative $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ images (in transverse orientation and relative scale) based on a series of 3D CSI data sets acquired with multiple RF pulse voltage values. The voxel-based $\mathbf{B}_1$ field images can be regressed and determined using the following equations [15]:

$$S \propto \rho \cdot \mathbf{B}_1^- \cdot SF \cdot \sin(\alpha), \quad \alpha = \gamma \cdot \tau \cdot \mathbf{B}_1^+ \cdot V \quad (6)$$

$$SF = \frac{1 - \exp(-\frac{TR}{T_1})}{1 - \cos(\alpha) \cdot \exp(-\frac{TR}{T_1})} \quad (7)$$

where S is the MR imaging signal intensity, ρ is the spin density, SF is a saturation factor and it equals to 1 under fully relaxed condition for CSI acquisition. Based on the fact that the imaging signal depends sinusoidally on $\mathbf{B}_1^+$, and the maximum signal at $\alpha=90^\circ$ is proportional to and determines the (relative) $\mathbf{B}_1^-$ value. Both $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ images can be estimated using a nonlinear curve fitting algorithm. Data processing was performed using a custom-built program in MATLAB software (v14.0, MathWorks).

SNR calculation for generating ^{17}O or ^{23}Na MRSI SNR maps

SNR was calculated by the ratio between the maximum ^{17}O or ^{23}Na signal intensity as regressed at $\alpha=90^\circ$ and the corresponding spectral noise level determined from the standard deviation (SD) of the spectral baseline regions in the absence of resonance signal. The SNR values were calculated for all MRSI voxels to generate SNR maps.

Simulation of $\mathbf{E}$ field and normalized noise level

We conducted a simulation study in CST Microwave Studio using a finite integration technique (FIT, a variant of FDTD) in the transient solver to estimate the $\mathbf{E}$ fields according to Eq. 8 and the normalized noise level (NNL) according to Eq. 9 [25] for the 10.5T ^{17}O MRSI phantom study in the presence and absence of the BST-based tuHDC ceramic based on the configuration as shown in Fig. S1a. The phantom has the relative permittivity $\epsilon_r=80$ and conductivity of $\sigma=1.0$ S/m. The RF coil operation frequency and optimal permittivity were set to $\omega=60.6$ MHz and $\epsilon_{r,\text{optimal}}=6000$, respectively.

RF resonator quality factor measurement

There were two types of RF resonator quality factor (Q) measurement used in this study: one is to measure the RF coil resonance frequency (f_0) width (Δf) measured at 3 dB point from the resonance peak based on S_{11} measurement using a network analyzer, and another one is to measure the self-resonance frequency width of the BST-based tuHDC block using a pick-up coil and S_{21} measurement (10 dB) [39, 40]. The Q value can be calculated according to $Q=f_0/\Delta f$.

III. RESULTS

Characteristic properties of the BST-based tuHDC ceramics for MR imaging applications

For a proof of concept, composite BST compounds were employed to construct low-tan δ tuHDC ceramic blocks (circular shape: 8 cm diameter and 2.1 cm thickness) for various MR imaging applications. Depending on the barium to strontium ratio (x , $0 < x < 1$) in the $\text{Ba}_x\text{Sr}_{1-x}\text{TiO}_3$ formulation, BST-ceramics can have an extremely high permittivity up to 10,000 at room temperature [34]. The ratio also determines the material properties of ϵ_r , tan δ and Curie temperature where the BST-ceramics reach the maximum permittivity and experience a phase change from the paraelectric to ferroelectric state. In this study, we employed the BST-based tuHDC ceramics ($\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$, $x=0.6$) for ^1H MRI application at 1.5T clinical scanner and two X-nuclear (^{23}Na and ^{17}O) MRSI applications at 7T and 10.5T human scanner, respectively, at RF operation frequency below 100 MHz.

Figures 1a and 1b show the experimentally measured ϵ_r and tan δ dependences on the temperature of BST-based tuHDC ceramic block (T_b). When T_b approached to the Curie temperature ($T_c \approx 0^\circ\text{C}$), ϵ_r reached a maximum (~ 16000), then dropped to ~ 4500 at T_{Room} and to ~ 2200 at $\sim 40^\circ\text{C}$ (Fig. 1a), exhibiting a very large tunable permittivity range. In parallel, tan δ also reached a maximum at the Curie temperature, and became much smaller at $10\text{--}30^\circ\text{C}$ (tan $\delta < 0.025$, Fig. 1b) in the paraelectric state (Fig. 1a), ideally for MR imaging applications.

When placing the BST-based tuHDC ceramic block at the center of a ^{17}O RF surface coil (Fig. S1a), the interaction between them shifted the RF coil resonant frequency (ω). Figure 1c shows the relationship between T_b and ω , which was originally set to 60.6 MHz (i.e., ^{17}O operation frequency at 10.5T) at 22°C ; and ω shifted towards a lower frequency when T_b decreased (without re-tuning the ^{17}O RF coil frequency). Figure 1d shows an inverse relationship between T_b and ϵ_r as calculated using Eq. 4. From Figs. 1c–1d, a linear relationship between ω and ϵ_r for the T_b range of $13\text{--}21^\circ\text{C}$ was derived (Fig. 1e). Therefore, the ϵ_r value or change can be empirically determined from T_b or ω shift ($\Delta\omega$).

The BST-based tuHDC ceramics had at least two self-resonance modes below 100 MHz and their self-resonant frequencies were temperature dependent (Figs. 2a–2b). When $T_b = T_{\text{Room}}$, the self-resonant frequencies of both modes were higher than the ^{17}O operation frequency at 10.5T (60.6 MHz). However, at lower T_b , the self-resonant frequencies approached the ^{17}O operation frequency around 12°C for the low-frequency mode and 4°C for the high-frequency mode, respectively, which can lead to passive coupling between the ^{17}O RF coil and tuHDC ceramic block. To avoid the coupling, we performed the ^{17}O MRSI

experiments at $T_b = 12^\circ\text{C}$ in this study. Figure 2c shows a linear relationship between the low-frequency mode self-resonance quality factor (Q) and T_b in the T_b range of 12–21 $^\circ\text{C}$. A similar trend between T_b and the ^{17}O RF coil Q was also observed (Fig. 2d), suggesting a strong interaction between the RF coil and tuHDC ceramic block, especially at lower T_b with a ultrahigh ϵ_r value as shown in Fig. 1.

Low-loss tuHDC ceramics improve $\mathbf{B}_1$ fields and reduce noise for ^{17}O MRSI at 10.5T

To evaluate the BST-based tuHDC ceramic performance, we conducted a ^{17}O MRSI study using a water phantom and a ^{17}O RF surface coil on 10.5T human scanner (see Fig. S1a). Figure 3a shows the representative ^{17}O spectra with maximum natural abundance ^{17}O -water signal in a central voxel of the three-dimensional (3D) ^{17}O MRSI acquired at $\alpha=90^\circ$, which was determined *via* the regression of the ^{17}O signal intensity versus the RF pulse voltage (Fig. 3b), under three conditions: control condition at 20 $^\circ\text{C}$ room temperature without the use of tuHDC ceramic block, and with the use of tuHDC ceramic block at $T_b=20^\circ\text{C}$ and at $T_b=12^\circ\text{C}$, respectively. First, the RF pulse voltage to reach $\alpha=90^\circ$ (i.e., V_{90°) was reduced by $\sim 50\%$, indicating approximately 4 times reduction in the required RF pulse power when using the tuHDC ceramic block at $T_b = T_{\text{Room}}$ and V_{90° was further reduced at $T_b = 12^\circ\text{C}$ (Fig. 3b). The results suggest large improvement of $\mathbf{B}_1^+$, which is inversely proportional to V_{90° , resulting in higher RF transmission efficiency. Second, the maximum ^{17}O signal intensity at $\alpha=90^\circ$ (i.e., S_{90°) increased significantly by using the tuHDC ceramic block (Fig. 3a), indicating an improved $\mathbf{B}_1^-$ field and SNR ($\propto \mathbf{B}_1^-$). Third, we discovered a large denoising effect by using the low-loss BST-based tuHDC ceramic block, especially at $T_b = 12^\circ\text{C}$ (Fig. 3a). Representative SNR images (transverse slice across the phantom center) as shown in Fig. 3c clearly demonstrate large SNR improvement at $T_b=12^\circ\text{C}$ as compared to control. Figure 3d shows a SNR gain for $>200\%$ in the object region near the tuHDC ceramic block at $T_b = 12^\circ\text{C}$ for the ^{17}O MRSI at 10.5T compared to control, though the improvement degree was reduced at $T_b = 20^\circ\text{C}$.

Figure 4a shows representative spectral noise images of the 3D ^{17}O MRSI datasets acquired with the BST-based tuHDC ceramic block at varied T_b compared to control from another experiment. The use of the tuHDC ceramic block resulted in a large global noise reduction across the entire imaging field of view (FOV). The denoising effect depended on T_b and associated ϵ_r , and reached best performance (55% noise reduction) at $T_b=15^\circ\text{C}$ (Figs. 4a–4b), thus, doubled the ^{17}O MRSI SNR.

Figure 4 also shows the $\mathbf{B}_1^+$, $\mathbf{B}_1^-$, noise level and SNR images from the ^{17}O MRSI datasets acquired with the tuHDC ceramic block at $T_b=15^\circ\text{C}$ (Fig. 4c), without the use of tuHDC (control, Fig. 4d) and their ratio images (Fig. 4e), as well as the 1D profiles along the vertical (Fig. 4f) and horizontal (Fig. 4g) dash lines as depicted in Fig. 4e. Although the improvements were spatially dependent (except the global denoising effect), the maximum and $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ were more than doubled, while the noise level reduced by $>50\%$, thereby increased SNR near the tuHDC ceramic block by $>300\%$ (Fig. 4g). Based on Fig. 1d, the optimal ϵ_r ($\epsilon_{r,\text{optimal}}$) for achieving the best performance at $T_b = 15^\circ\text{C}$ was extremely high (≈ 6030) for the 10.5T ^{17}O MRSI application at $\omega=60.6\text{ MHz}$.

Low-loss tuHDC ceramics improve B_1 fields and reduce noise for ^{23}Na imaging at 7T

Sodium (^{23}Na) MRI provides critical information of brain pathophysiology, especially at UHF [41]. In this study, we tested whether the same BST-based tuHDC ceramics can be employed to improve ^{23}Na MRSI at 7T ($\omega=78.6$ MHz) and achieve optimal performance at room temperature, thus, a temperature controller is not needed. Figure 5a displays the ^{23}Na MRSI noise images acquired without (control) and with the tuHDC ceramic block at varied T_b , and the corresponding noise levels of the ^{23}Na MRSI voxels are shown in Fig. 5b, indicating a substantial global denoising effect ($\sim 10\%$) at $T_b \approx 25$ °C.

Figure 5c shows the T_b -dependent B_1^+ , B_1^- and SNR measured in the central ^{23}Na MRSI voxel (depicted in Fig. 5f), collectively indicating an optimal condition for best performance at $T_b = 20$ °C, i.e., at T_{Room} . The B_1^+ , B_1^- and SNR images acquired at T_{Room} in the presence (Fig. 5d) and absence (Fig. 5e, control) of the tuHDC ceramic block, and their ratio images (Fig. 5f) are displayed. The overall results indicate an $>200\%$ B_1^+ and B_1^- improvements, and $>300\%$ SNR improvement in the sample region near the ceramic block at T_{Room} . The $\epsilon_{r,\text{optimal}}$ value for 7T ^{23}Na MRSI application ($\omega=78.6$ MHz) at room temperature was ~ 4500 .

Low-loss tuHDC ceramics improve B_1 fields and detection sensitivity for ^1H MRI on a 1.5T clinical scanner

We hypothesize that the performance of the tuHDC-RF-coil technology is determined by the MR imaging operation (or Larmor) frequency and corresponding $\epsilon_{r,\text{optimal}}$, thus, the same BST-based tuHDC ceramics could be applied to ^1H MRI at a lower field strength operated at <100 MHz. Since the γ ratio between ^1H and ^{17}O nuclides is approximately 7, the ^1H MRI operation frequency at 1.5T (64 MHz) is very close to that of ^{17}O MRSI at 10.5T (60.6 MHz); therefore, we examined whether ^1H MRI on a 1.5T clinical scanner could also benefit significantly from the tuHDC-RF-coil technology.

Figure 6a shows representative ^1H water signal images acquired using the BST-based tuHDC ceramic block with varied T_b between 12 to 20 °C as compared to control. A large MRI signal enhancement was evident at $T_b < T_{\text{Room}}$, and it reached a maximum at $T_b \approx 14$ °C, corresponding to $\epsilon_{r,\text{optimal}} \approx 6400$. Figures 6b and 6c show the B_1^+ and B_1^- ratio images between with and without the use of tuHDC ceramic block at varied T_b , which were highly T_b -dependent and reached best performance at T_b of ~ 14 °C. The 1D B_1^+ and B_1^- ratio profiles reveal more than 200% B_1^+ enhancement (Fig. 6b); and strikingly the B_1^- ($\propto$ SNR) increased over 400–500% near the tuHDC ceramic block (Fig. 6c). The optimal values of $\epsilon_{r,\text{optimal}} \approx 6400$ and $T_b \approx 14$ °C for 1.5T ^1H MRI application are similar to the 10.5T ^{17}O MRSI results ($\epsilon_{r,\text{optimal}} \approx 6030$ and $T_b \approx 15$ °C), since their RF operation frequencies are close. These findings confirm that the key factor in determining optimal ϵ_r and T_b is the operation frequency of a desired nuclide MR application at a given B_0 .

Low-loss non-tunable uHDC ceramic block reduces noise for ^{31}P MRSI of phantom and human brain at 7T

The large denoising effect as found in the 7T ^{23}Na MRSI and 10.5T ^{17}O MRSI studies was critical for gaining SNR. Such denoising effect was also observed in the ^{31}P MRSI phantom and human brain studies at 7T ($\omega=120$ MHz) using a curved low-loss uHDC ceramic block made of $\text{Pb}(\text{Zr}_{0.5}\text{Ti}_{0.5})\text{TiO}_3$ ($\tan\delta \approx 0.05$) with a temperature-independent ϵ_r value (≈ 1200). We observed approximately 43% and 16% noise reduction in the phantom (Fig. 7a) and human brain (Fig. 7b), respectively. In the ^{31}P MRSI phantom study, the large FOV was intentionally set to cover not only the phantom, but also outside the RF head volume coil (Fig. S1b). Interestingly, the noise reduction appeared everywhere in image space and we also observed the denoising effect ($\sim 20\%$) in the absence of the phantom (unloaded condition, Fig. S1c). Figure 7c shows *in vivo* ^{31}P MRSI results of healthy human brain at 7T, indicating that the application of the curved low-loss uHDC ceramic block improved SNR for $>100\%$ and reduced the required RF pulse power for $>85\%$ for human brain application.

In another 7T ^{31}P MRS phantom study, we used a higher dielectric loss uHDC ceramic block made of ferroelectric lead zirconate titanate (PZT) material with a Q value 6 times lower than that of the BST-ceramic blocks [30]. Although both $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ were significantly enhanced, the noise level was also increased, thus, reduced the degree of improvement. Therefore, low $\tan\delta$ is a critical feature of the uHDC/tuHDC materials for achieving a large noise reduction and high SNR gain.

IV. DISCUSSION

In this work, we introduced a novel tuHDC-RF-coil technology that integrates RF coils with a highly ϵ_r tunable (from ~ 2000 to >15000 for the T_b range of $0-40^\circ\text{C}$) and low-loss ($\tan\delta < 0.03$, for $10-30^\circ\text{C}$ T_b) tuHDC ceramics based on the BST material (Figs. 1a–1b) to optimize and greatly improve the performance for broad MR imaging applications covering different field strengths and nuclides of biological significance.

The tuHDC-RF-coil technology significantly increases the $\mathbf{B}_1^+$ field, thereby reduces the required RF pulse power. For instance, we observed $>200 \mathbf{B}_1^+$ improvement for ^{23}Na MRSI at 7T (Figs. 5c–5f) and ^1H MRI at 1.5T (Fig. 6b), thus, reducing RF pulse power by approximately 90% for achieving the same RF pulse flip angle. Synergistically, it produces a much stronger $\mathbf{B}_1^-$ field for various MR imaging applications (Figs. 3b and 4 for ^{17}O MRSI at 10.5T, Fig. 5 for ^{23}Na MRSI at 7T, Fig. 6c for ^1H MRI at 1.5T, and Fig. 7c for ^{31}P MRSI at 7T), resulting in higher detection sensitivity ($\propto \mathbf{B}_1^-$), thus, a larger SNR gain.

Strikingly, we discovered that the tuHDC-RF-coil (or uHDC-RF-coil) technology produces a large and global denoising effect of $>50\%$ for ^{17}O MRSI at 10.5T (Fig. 4b), $\sim 10\%$ for ^{23}Na MRSI at 7T (Fig. 5b), and 43% (phantom, Fig. 7a) and 16% (human brain, Fig. 7b) for ^{31}P MRSI at 7T under optimal condition. The large denoising effect observed in this study cannot be explained by the traditional RF coil theory that links RF coil Q and noise. In the

10.5T ^{17}O MRSI study, the maximum denoising effect was observed at $T_b = 14^\circ\text{C}$, while the ^{17}O RF coil Q (≈ 35) was decreased 35% from the condition at $T_b = 20^\circ\text{C}$ ($Q \approx 54$, Fig. 2d). To understand the mechanism underlying the denoising phenomenon, we investigated the relationship between MR imaging noise and $\mathbf{E}$ field, which consists of conservative ($\mathbf{E}_C$) and non-conservative ($\mathbf{E}_{NC}$) components according to Eq. 8,

$$\mathbf{E} = \overbrace{-\nabla\phi}^{\text{conservative}} - \overbrace{i\omega\mathbf{A}}^{\text{non-conservative}} \quad (8)$$

where ϕ is the electric scalar potential, $\mathbf{A}$ is the magnetic vector potential. In the near field condition, the $\mathbf{E}_C$ term in the sample contributes to the noise significantly. In addition to the enhancement of displacement current as a $\mathbf{B}_1$ field source, the tuHDC ceramics also partially shield $\mathbf{E}_C$ of the RF coil, which consequently, reduce the normalized noise level (NNL) from the sample according to Eq. 9,

$$\text{NNL} = \sqrt{\frac{\int \sigma |\mathbf{E}_C|^2 dv}{\text{avg}(\mathbf{B}_1^2)}}, \quad (9)$$

and lead to the denoising effect as observed in this study. We conducted a computer simulation of the ^{17}O MRSI phantom study at 10.5T (Fig. S1a) using $\epsilon_{r,\text{optimal}} \ll 6000$. The simulation shows that the tuHDC ceramic block reduced $\mathbf{E}$ field (Fig. 8a) and enhanced $\mathbf{B}_1^+$ field (Fig. 8b), which partially contributed to the large reduction of $\mathbf{E}_C$ field (normalized by the average $\mathbf{B}_1^+$ in the sample, Fig. 8c). The reduced $\mathbf{E}_C$ field accompanying with increased $\mathbf{B}_1^+$ results in a large NNL reduction of 62% based on the simulation result, which is consistent with the experimentally detected noise reduction of 55% in this study (Fig. 8d).

The tuHDC-RF-coil technology provides an unprecedented SNR improvement. For example, the best SNR improvement of 10.5T ^{17}O MRSI near the tuHDC ceramic block reached 4 to 5 folds (Fig. 4) owing to the large denoising effect and $\mathbf{B}_1^+$ enhancement. Based on the field dependence of ^{17}O MRSI SNR ($\propto B_0^{7/4}$), such a high SNR achieved at 10.5T could reach a similar performance on a MRI scanner at 23T. The large $\mathbf{B}_1^+$ or SNR improvement (5–6 times) observed in the 1.5T ^1H MRI study in the object near the tuHDC ceramic block (Fig. 6c) is equivalent to either using less than 1/25 scan time to achieve the same MRI quality; or a similar ^1H MRI SNR ($\propto B_0$) at UHF of above 7T.

The tuHDC-RF-coil technology for MR imaging applications operated at <100 MHz requires the use of an ultrahigh permittivity material, which generates a large displacement current and reduces the conservative $\mathbf{E}$ field for a RF coil; therefore, significantly enhances the $\mathbf{B}_1^+$ and $\mathbf{B}_1^-$ fields and induces a large noise reduction, leading to an unprecedented SNR gain.

The similar improvements between the 1.5T ^1H MRI and 10.5T ^{17}O MRSI studies confirm that the performance of the tuHDC-RF-coil technology depends on the RF operation frequency and the corresponding $\epsilon_{r,\text{optimal}}$, which can be robustly optimized by varying T_b .

For the same reason, the large improvement observed for ^{23}Na MRSI at 7T and room temperature in this study could be translatable to *in vivo* ^{13}C MRSI application at the same field strength because of their similar RF operation frequencies.

Figure 9 summarizes the regressed relationship between the RF operation frequency and the optimal permittivity of HDC/uHDC/tuHDC materials for a variety of nuclides at varied field strengths. The non-linear regression of the curve shown in Fig. 9 provides a relationship between the MR imaging operation frequency (ω , in MHz unit) of a desired nuclide and the optimal permittivity ($\epsilon_{r,\text{optimal}}$) of the BST-based tuHDC ceramic block ($x=0.6$) according to the regressed Eq. 10:

$$\epsilon_{r,\text{optimal}} \approx \frac{7 \times 10^7}{\omega^{2.248}}. \quad (10)$$

The relationship between the optimal temperature ($T_{b,\text{optimal}}$) of the BST-based tuHDC ceramics and ω can be derived from Eqs. 4 and 10,

$$T_{b,\text{optimal}} = 1.29 \times 10^{-3} \times \omega^{2.248}. \quad (11)$$

The estimated $\epsilon_{r,\text{optimal}}$ value for ^1H MRI application at UHF of 7T (or operated at ω 298 MHz) will be much lower, for instance, being ~ 77 for ^1H MRI/MRS(I) at 10.5T, thus, the enhancement of the displacement current as well as improvements in $\mathbf{B}_1^+$, $\mathbf{B}_1^-$ and SNR are anticipated to be moderate. In contrast, the application of ^1H MRI at 1T or ^2H MRSI at 7T operated at 43 or 46 MHz requires an extremely high $\epsilon_{r,\text{optimal}}$ of ~ 15000 or 13000 , respectively, according to Fig. 9 Eq. 10, which could be reachable for the BST-based tuHDC ceramic blocks when T_b drops toward the Curie temperature ($T_c \approx 0^\circ\text{C}$, Fig. 1a). For instance, the optimal T_b for ^2H MRSI application at 7T ($\omega=45.8$ MHz) was estimated to be approximately 7°C according to Eq. 11, which corresponds to $\epsilon_{r,\text{optimal}} = 12957$; interestingly, without a passive coupling between the ^2H RF coil and tuHDC self-resonance frequencies (~ 6 MHz apart, Figs. 2a–2b). Equations 10 and 11 are valuable to design and optimize the BST-based tuHDC-RF-coil technology for broad MR imaging applications as demonstrated in this work.

The BST-based tuHDC ceramic materials and performance for MR imaging applications can be further improved by tailoring the material property *via* adjusting the chemical composition fraction x of $\text{Ba}_x\text{Sr}_{1-x}\text{TiO}_3$. Increasing x from 0.6 towards 0.7 will shift the T_b -dependent permittivity curves towards a higher temperature as schematically illustrated in Fig. S2. This property provides a unique strategy for achieving an ultrahigh $\epsilon_{r,\text{optimal}}$ for a particular MR imaging application of interest at a higher T_b , for instance, at room temperature by optimizing the x value. The 7T ^{23}Na MRSI study using $x=0.6$ as demonstrated herein proves the concept. Further increase in the x value can substantially boost the tuHDC ceramic permittivity to > 6000 at room temperature, thus, potentially for optimal 1.5T ^1H MRI or 10.5T ^{17}O MRSI application at $T_b \approx T_{\text{Room}}$, thus, no temperature controller is needed. On the other hand, decreasing x can reduce the permittivity to 1000–

2000 at room temperature, thus, for optimal ^{31}P MRSI application at 7T or ^1H MRI application at 3T as guided by Fig. 9.

Variation in the tuHDC ceramic geometry might slightly change the relationship between ω and optimal ϵ_r or T_b , nevertheless, the optimal condition for a tuHDC ceramic application with an arbitrary design, size or geometry of interest can always be calibrated and optimized *via* the similar experimental procedure as demonstrated herein. Finally, the majority of this work was conducted using one BST-based tuHDC ceramic block for the proof of concept for several MR imaging applications, the same tuHDC-RF-coil technology can be readily extended to multiple tuHDC blocks or human head-shape tuHDC helmet (for instance) integrated with RF array or volume coils.

V. CONCLUSION

This work introduces an innovative RF engineering solution by incorporating the BST-based tuHDC ceramics with a RF coil, leading to an unprecedented SNR gain for a variety of MRI/MRSI applications at field strengths ranging from 1.5T to 10.5T. This technology is robust and cost-effective, and is highly valuable for broad MRI and MRS(I) applications. It presents a technological breakthrough for pushing sensitivity and spatiotemporal resolution limits beyond the current state-of-the-art MR imaging technology; therefore, it could significantly advance basic and clinical research, and more importantly improve the capability of diagnostic imaging. The same concept can be extended to other tuHDC materials beyond the BST-based ceramics and to organs other than the brain as investigated in this study.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1. Temperature dependence of dielectric permittivity and loss tangent of the $\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$ (BST) ceramic block, and the relationship between the permittivity and RF coil resonant frequency.

Experimental results of (a) the permittivity and (b) dielectric loss $\tan \delta$ of the BST ($x=0.6$) ceramic block over a temperature range of 40 to -40°C at 10 kHz and 100 MHz operation frequency, respectively; and (c) the relationship between the RF coil resonant frequency and the tuHDC block temperature. The RF coil frequency was first tuned and matched to the ^{17}O operation frequency (60.6 MHz at 10.5T) at $T_b = 22^{\circ}\text{C}$, and was not re-tuned or re-matched when the tuHDC block temperature was below 22°C . (d) Relationship between the BST ceramic block temperature and permittivity estimated based on Eq. 6 with the parameters of $T_c \approx 0^{\circ}\text{C}$ and $\epsilon_r \approx 4500$ at the room temperature of 20°C . (e) A linear correlation between RF coil resonance frequency and the BST-based tuHDC permittivity within the ϵ_r range of 4000 – 7000 . The regression slope indicates that the RF coil frequency increase of 1 MHz corresponds to a decrease in ϵ_r of 504 for the BST-based tuHDC ceramic block.

Figure 2. Self-resonant frequency and quality Q of $\text{Ba}_{0.6}\text{Sr}_{0.4}\text{TiO}_3$ (BST) ceramic blocks and their temperature dependence.

(a, b) Temperature-dependent self-resonant frequency of two representative BST-based tuHDC ceramics, respectively, at two resonance modes (low versus high frequency) measured with a pickup probe using a network analyzer, (c) Temperature-dependent self-resonant quality factor Q of the BST-based tuHDC ceramic block (low-frequency mode), (d) Relationship between the ^{17}O RF coil quality factor Q (for 10.5T application) and the tuHDC ceramic temperature when the coil was integrated with the tuHDC ceramic block.

Figure 3. Experimental results demonstrate significant improvement for the ^{17}O MRSI application using the BST-based tuHDC ceramic block at 10.5T.

(a) ^{17}O spectra of nature abundance ^{17}O -water signal measured from a central voxel of 3D ^{17}O MRSI data acquired at 10.5T from a water phantom with an optimal RF pulse voltage for reaching nominal 90° RF pulse flip angle; (b) the maximum water ^{17}O signal was determined *via* regression of the voxel ^{17}O signal intensity versus the RF pulse voltage under three conditions: control without tuHDC at the room temperature (20°C , black lines and symbols), with the tuHDC ceramic block at $T_b = 20^\circ\text{C}$ (blue lines and symbols) and at $T_b = 12^\circ\text{C}$ (red lines and symbols), respectively; (c) the representative SNR images (transverse slices across the center of the cylindrical phantom) acquired under the three conditions; and (d) the SNR ratio Image between the use of tuHDC ceramic block at $T_b = 12^\circ\text{C}$ and the control condition without the tuHDC ceramic block. The box in c and d depicts the ^{17}O MRSI voxel location employed in a and b.

Figure 4. Significant enhancement of RF coil transmission and reception efficiency, large denoising effect and SNR improvement for 10.5T ^{17}O MRSI application using the BST-based tuHDC ceramic block.

(a) Representative 3D ^{17}O MRSI noise images acquired with the BST-based tuHDC ceramic block at varied temperature (T_b) as compared to the control condition without the ceramic block at room temperature (20 °C), and (b) ^{17}O MRSI voxel-wise noise levels from 550 ^{17}O MRSI voxels and their averaged noise levels (yellow lines; in arbitrary unit: A.U.) acquired with the tuHDC ceramic block with varied T_b (red lines and symbols) and under control (blue lines and symbols). Representative images (in transverse orientation) of B_1^+ , B_1^- , noise level and SNR measured in the presence of the BST-based tuHDC ceramic block at $T_b=15$ °C (c) and control condition without the use of the ceramic block (d), and their ratio images (e); and corresponding 1D profiles taken from the B_1^+ , B_1^- and SNR ratio images in the horizontal (X axis) and vertical (Y axis) directions (along the dashed lines displayed in (e)) are shown in (f) and (g), respectively.

Figure 5. Significant improvement of RF coil transmission and reception efficiency, substantial denoising effect and large SNR gain for 7T ^{23}Na MRSI application using the BST-based tuHDC ceramic block.

(a) Representative 3D ^{23}Na MRSI noise images acquired with the BST-based tuHDC ceramic block at varied block temperature (T_b) as compared to the control condition without the use of the tuHDC ceramic block at room temperature (20 °C). (b) ^{23}Na MRSI voxel-wise noise levels from 550 voxels and their averaged noise levels (yellow lines; in arbitrary unit: A.U.) acquired with the tuHDC ceramic block with varied T_b (red lines and symbols) and under control (blue lines and symbols), respectively, (c) Temperature dependences of B_1^+ , B_1^- and SNR based on the measurement in a central voxel of 3D ^{23}Na MRSI showing an optimal performance at near room temperature of ~20 °C. Representative B_1^+ (top) and B_1^- (middle) and SNR (bottom) images (in transverse orientation) obtained using the BST-based tuHDC ceramic block at room temperature (d) as compared to the control condition without the use of the tuHDC ceramic block (e), and their ratio images (f). The red circles in (c) point to the T_b values resulting in the best performance.

Figure 6. Large improvement of ^1H RF coil transmission efficiency and MRI detection sensitivity on a 1.5T clinical scanner using the BST-based tuHDC ceramic block.

(a) Representative ^1H MRI water signal intensity images (central transverse slice) acquired with the same RF pulse voltage using the BST-based tuHDC ceramic block at varied T_b as compared to the control condition without the use of the tuHDC ceramic block at room temperature ($\sim 20^\circ\text{C}$). Temperature dependence of representative B_1^+ (b) and B_1^- (c) ratio images and their 1D ratio profiles (Y axis) measured using the tuHDC ceramic block at the optimal T_b of $\sim 14^\circ\text{C}$ versus the control condition.

Figure 7. Significant denoising effect and SNR gain observed in the 7T ^{31}P MRSI studies using a low-loss curved uHDC ceramic block with a fixed ultrahigh permittivity (≈ 1200).

(a) ^{31}P MRSI noise maps acquired from a Pi phantom study using a ^1H - ^{31}P dual frequency TEM head coil in the absence (control) and presence of the uHDC ceramic block (blue bar), as well as the 3D ^{31}P MRSI voxel-based noise level comparison between the two conditions averaged over 550 voxels for each condition, (b) *In vivo* ^{31}P MRSI noise maps and noise level comparison obtained from a human brain at 7T using the same ^1H - ^{31}P TEM head coil in the absence (control) and presence of the same uHDC block at room temperature, (c) *In vivo* ^{31}P MRSI results obtained with (left, RF pulse voltage $V=50$ v) and without (right, $V=150$ v) using the same TEM head coil and uHDC ceramic block placed under the occipital lobe of human brain at 7T. The ^{31}P CSI spectra demonstrate >100% SNR gain using the uHDC technique and only 1/3 RF pulse voltage or <15% RF power was used as compared to that without the use of the uHDC block.

Figure 8. Computer simulation results and comparison of NNL, E and B_1^+ fields for the ^{17}O MRSI phantom application at 10.5T in the absence (control) and presence of the BST-based tuHDC ceramic block at 15 °C with $\epsilon_{r,\text{optimal}}$ of 6000.

(a) Normalized E-field maps inside and outside the phantom (outlined by cycles), and (b) B_1^+ field maps in control (left panels) and presence of the tuHDC ceramic (right panels), (c) The profile of normalized E-field divided by the average B_1^+ over the space (inside the phantom) along the vertical dashed lines in (a), (d) The simulation and experimental NNL results in the absence (control) and presence of the tuHDC ceramic block for comparison.

Figure 9. The relationship between the optimal permittivity and the magnetic resonance operation frequency for various nuclides or field strengths.

A diagram illustrating the relationship between the optimal permittivity ($\epsilon_{r,optimal}$) of HDC, uHDC or tuHDC materials and the MR operation frequency (ω), covering a wide range of MR imaging applications for various nuclides either at the same or different magnetic field strengths; as well as the empirical regression equation describing this relationship based on the experimental results (blue full cycles). Red circles present the estimated $\epsilon_{r,optimal}$ values for the nuclides of interest at a given field strength.