

Simulation Study of Low-Field Proton Magnetic Resonance Spectroscopy Using Optically Pumped Magnetometers

Ryo Enari,¹ Hiroyuki Ueda,¹ Kazuyoshi Yoshii,¹ and Yosuke Ito¹

Department of Electrical Engineering, Graduate School of Engineering, Kyoto University, Kyoto, Japan

(*Electronic mail: ryo.enari.22s@st.kyoto-u.ac.jp)

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We numerically investigated the feasibility of low-field magnetic resonance spectroscopy (MRS) using optically pumped magnetometers (OPMs). Low magnetic fields reduce signal strength and spectral resolution, challenging reliable metabolite quantification. To address these issues, we performed Bloch-equation-based simulations at 0.01–7 T using aqueous phantoms. We compared chemical shift selective (CHESS) and inversion recovery (IR) methods for water suppression. We also evaluated signal detection using both radio-frequency (RF) coils and OPMs. The IR method, relying on T_1 contrast rather than frequency selectivity, effectively suppressed water signals down to 0.1 T. The signal-to-noise ratio (SNR) of Glu increased below 0.7 T. OPMs maintained nearly constant SNR across 0.01–1 T and outperformed coils below 1 T. In mixed-metabolite settings, however, spectral overlap and IR-induced co-suppression impaired detection, particularly for Gln below 0.1 T. These results show that IR and OPMs offer complementary advantages for low-field MRS, while spectral resolution remains a challenge. This work provides a theoretical basis for enhancing low-field MRS techniques and expanding access to metabolic diagnostics in clinical and research settings.

Proton magnetic resonance spectroscopy (^1H -MRS) provides non-invasive, molecular-level insights for diagnosing neurological and cerebrovascular diseases^{1–5}. While conventional high-field MRS yields high signal-to-noise ratio (SNR) and spectral resolution, it is technically demanding and requires specialized, expensive infrastructure. This limitation has promoted growing interest in low-field MR approaches that promise to broaden the reach of MR-based techniques, particularly in resource-limited settings^{6–8}.

Among low-field MR techniques, low-field magnetic resonance imaging (MRI) has advanced rapidly in recent years owing to recent advances in artificial intelligence^{8–10}. However, the potential of MRS under similar conditions has not been extensively explored¹¹. This limited progress mainly stems from unresolved physical and technical challenges associated with decreasing Larmor frequencies. At low magnetic fields, reduced chemical shift dispersion and signal sensitivity degrade spectral resolution and quantification of metabolites.

First, the chemical shift between water and metabolite peaks narrows, reducing the effectiveness of frequency-selective suppression methods such as CHESS¹² and VAPOR¹³, since RF pulses lack sufficient selectivity at low fields. Second, the reduced Larmor frequency weakens the signal amplitude detected by conventional RF coils. Although these effects have been primarily characterized in the context of NMR spectroscopy and low-field MRI^{14,15}, the same physical constraints fundamentally limit the sensitivity of MRS, causing a marked decline in SNR, especially in the sub-tesla regime. These limitations compromise the reliability of metabolite quantification and must be overcome for the practical implementation of low-field MRS.

To address the problem of water suppression at low fields, we introduce inversion recovery (IR)¹⁶ as an alternative to frequency-selective techniques. IR leverages the difference in T_1 relaxation times between water and metabolites, rather than their frequency separation, enabling suppression performance that is independent of chemical shift. This makes IR

especially suitable for low-field environments, in which conventional techniques like CHESS fail to provide adequate selectivity. By employing IR, we aim to evaluate and compare water suppression strategies that remain effective even under conditions of reduced spectral resolution.

To compensate for the degraded sensitivity at low fields, we adopt optically pumped magnetometers (OPMs)^{17,18} in place of traditional RF coils. OPMs detect magnetic fields via quantum spin polarization in alkali vapor, allowing them to maintain high sensitivity in low fields. They also eliminate the need for cryogenic cooling and offer compact, flexible system configurations. Given these advantages, OPMs are a promising detection technology for enhancing the performance of low-field MRS¹⁹.

To address reduced spectral resolution and sensitivity at weak magnetic fields, we simulated a low-field MRS system integrating IR-based water suppression and OPM-based detection. Simulations were conducted across 0.01–7 T. We quantified SNR under single- and mixed-metabolite conditions and identified limitations from spectral overlap and relaxation. These results provide a theoretical basis for extending MRS to environments lacking high-field infrastructure.

In ^1H -MRS, electron shielding alters the effective magnetic field depending on the local chemical environment, producing chemical shifts δ (in ppm relative to tetramethylsilane). These shifts enable the identification of metabolites like N-acetylaspartate (NAA), glutamate (Glu), and glutamine (Gln). Within the Bloch-equation framework, chemical shifts were implemented as frequency offsets in the B_z component. Accordingly, the resonance offsets of NAA, Glu, and Gln were assigned to 2.02, 2.35, and 2.45 ppm, respectively, and scaled linearly with the static field strength B_0 .

Figure 1 illustrates the proposed OPM-based MRS system. Figure 1 (a) shows the overall configuration, in which the MR signal generated in the sample is transferred to the OPM sensor via a flux transformer (FT)^{20,21}. Because the gyromagnetic ratios of protons and electrons differ by three orders of magni-

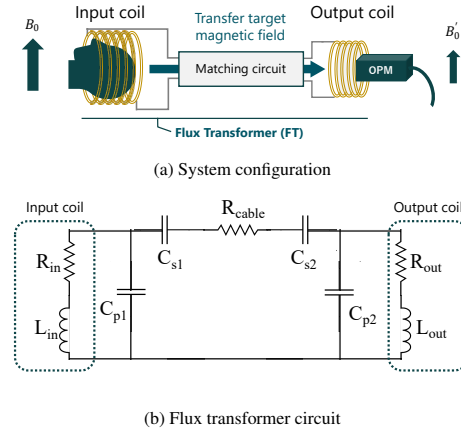


FIG. 1: Overview of the OPM-MRS system and flux transformer (FT) circuit model. (a) Schematic of the proposed system, where the MR signal is transferred from the sample to the sensor via an FT for OPM detection. (b) Equivalent circuit model of the FT, consisting of an input coil, output coil, matching capacitors, and cable resistance. This configuration enables resonant signal transfer while isolating the sample and detection regions.

tude, the Larmor frequency of the MR signal falls far outside the operational range of OPMs under the same static field.

To overcome this mismatch, the FT, configured as a resonant RLC circuit, transfers the MR signal from the sample to a detection region under a different magnetic field. This configuration bridges the frequency gap while improving detection through resonance selectivity. Compared with conventional inductive detection, it reduces coil resistance and associated thermal noise, improving sensitivity for low-field MRS. The subsequent circuit simulations for the FT and OPM detection followed this configuration. To ensure optimal signal transfer, the matching capacitors (C_p and C_s) on both the input and output sides were tuned to achieve resonance at each target Larmor frequency.

We implemented two water suppression strategies and evaluated them under identical acquisition settings: CHESS and IR. In CHESS, a three-pulse frequency-selective saturation module centered at the water resonance (4.70 ppm) was applied before excitation, with crusher gradients after each pulse. The pulse parameters were adjusted to suppress water while preserving nearby metabolite peaks.

In IR, a non-selective 180° inversion precedes excitation. The inversion time (TI) at each B_0 was determined from the water longitudinal relaxation time T_{1,H_2O} as $TI \approx T_{1,H_2O} \ln 2$ and refined in simulation to minimize residual water at readout. This approach exploits T_1 contrast rather than frequency selectivity. Except where noted, signal acquisition and analysis (including windowing and SNR estimation) were identical

between CHESS and IR conditions. Therefore, any observed differences in performance arose solely from the suppression method itself.

The dynamics of 1H nuclear magnetization was calculated using the Bloch equation²²:

$$\frac{d}{dt} \mathbf{M} = \gamma_p \mathbf{M} \times \mathbf{B} - \begin{bmatrix} \frac{M_x}{T_2}, \frac{M_y}{T_2}, \frac{M_z - M_\infty}{T_1} \end{bmatrix}^T$$

$$= \begin{bmatrix} -R_2 & \gamma_p B_z & -\gamma_p B_y \\ -\gamma_p B_z & -R_2 & \gamma_p B_x \\ \gamma_p B_y & -\gamma_p B_x & -R_1 \end{bmatrix} \begin{bmatrix} M_x \\ M_y \\ M_z \end{bmatrix} + \begin{bmatrix} 0 \\ 0 \\ R_1 M_\infty \end{bmatrix}, \quad (1)$$

where γ_p , $\mathbf{M}$, $\mathbf{B}$, $R_1 (= 1/T_1)$, $R_2 (= 1/T_2)$, and M_∞ denote the gyromagnetic ratio of proton spin, proton magnetization vector, magnetic field vector, longitudinal relaxation rate, transverse relaxation rate, and thermal equilibrium magnetization, respectively.

The values of M_∞ were analytically derived from the Boltzmann distribution based on an aqueous phantom volume of 250 mL, an absolute temperature of 286 K, and the respective molar concentrations of the spins. The target metabolites are NAA, Glu, and Gln, at concentrations of 10, 10, and 2 mM, respectively chosen for their biological relevance to neuronal and glial metabolism^{23–25}. The magnetization trajectories were computed by integrating the Bloch equations using a fourth-order RungeKutta method with a time step of $\Delta t = 2.5/2^{18}$ (9.5 μ s). All computations used double-precision (64-bit) arithmetic to ensure numerical stability. A simple free-induction decay (FID) acquisition with a duration of 2.5 s was assumed, and detector noise was incorporated at the signal evaluation stage according to each detection model. To ensure realism, the overall scaling factor of the simulated signal magnitudes was calibrated by normalizing water proton intensity to match the experimental SNR at 0.3 T.

We experimentally measured the water signal with a commercial 0.3 T MRI scanner. The measured values are listed in Table I. We confirmed their consistency with the Bloembergen-Purcell-Pound (BPP) relaxation theory²⁶ to extrapolate the relaxation times to other magnetic field strengths. For metabolites, the relaxation times of NAA were determined using empirical formulas reported for rat brain spectroscopy²⁷:

$$T_{1,NAA} = \alpha B_0^\beta \quad (2)$$

$$T_{2,NAA} = \gamma \exp(-B_0/\delta). \quad (3)$$

The empirical parameters for NAA were set to $\alpha = 1287.7$, $\beta = 0.118$, $\gamma = 459.65$, and $\delta = 23.9$.

To address the lack of established empirical formulas for Glu and Gln across all field strengths, their relaxation times were estimated by scaling the NAA values. We used constant scaling factors derived from established high-field literature values²⁸. Specifically, the ratios $T_{1,Glu}/T_{1,NAA} \approx 0.71$, $T_{1,Gln}/T_{1,NAA} \approx 0.70$, $T_{2,Glu}/T_{2,NAA} \approx 0.83$, and $T_{2,Gln}/T_{2,NAA} \approx 0.54$ were applied across all simulated magnetic fields. Table I shows a summary of the calculated relaxation times at 0.3 T for each metabolite.

Signal detection using the OPMFT configuration was modeled as follows. The MR signal B_{in} induces a voltage in the

TABLE I: Simulation parameters for each metabolite (reference values at 0.3 T).

Metabolite	Chemical shift (ppm)	T_1 (ms)	T_2 (ms)
H ₂ O	4.70	980	980
NAA	2.02	1171	454
Glu	2.35	786	246
Gln	2.45	793	376

input coil L_{in} (Fig. 1). The resulting current I_{out} through the output coil L_{out} generates an output magnetic flux density $B_{out} = L_{out}I_{out}/S_{out}$, where S_{out} is the cross-sectional area of the coil. B_{out} served as the input magnetic field for the OPM simulation described below:

$$\begin{aligned}
 \frac{d}{dt}\mathbf{S} &= \frac{\gamma_e}{q}\mathbf{S} \times \mathbf{B} - \left[\frac{S_x}{T_2^{(e)}}, \frac{S_y}{T_2^{(e)}}, \frac{S_z - S_\infty}{T_{op}} \right]^T \\
 &= \begin{bmatrix} -R_2^{(e)} & (\gamma_e/q)B_z & -(\gamma_e/q)B_y \\ -(\gamma_e/q)B_z & -R_2^{(e)} & (\gamma_e/q)B_x \\ (\gamma_e/q)B_y & -(\gamma_e/q)B_x & -R_{op} \end{bmatrix} \begin{bmatrix} S_x \\ S_y \\ S_z \end{bmatrix} \\
 &+ \begin{bmatrix} 0 \\ 0 \\ (R_{op}/q) \cdot S_\infty \end{bmatrix}.
 \end{aligned} \quad (4)$$

Here, $\gamma_e = 1.76 \times 10^{11} \text{ rad s}^{-1} \text{ T}^{-1}$ is the gyromagnetic ratio of bare electrons, $q = 4$ is the slowdown factor, $\mathbf{S}$ denotes the electron spin polarization vector, and $\mathbf{B}$ is the transmitted magnetic field vector applied to the OPM. $R_{op} = 2.0 \times 10^3 \text{ s}^{-1}$ and $R_2^{(e)} = 1.0 \times 10^3 \text{ s}^{-1}$ represent the optical pumping and transverse relaxation rates, respectively, while the thermal equilibrium polarization was set to $S_\infty = 0.5$.

The transmitted magnetic field was superimposed on a static bias field to form the vector $\mathbf{B}$. The frequency response of the OPM was macroscopically modeled as a transfer function. To ensure quantitative realism, the simulated signal magnitude was calibrated using experimental data acquired at 0.3 T, and the magnetic field sensitivity was determined by comparing this signal to the sensor's theoretical noise floor dominated by photon shot noise and spin-projection noise, while the chemical shift scale was referenced to the water resonance.

Noise sources were modeled for both the FT circuit and the OPM detector. To incorporate explicit stochastic elements, pseudo-random Gaussian white noise was generated with a standard deviation equal to the theoretical root-mean-square (RMS) noise amplitude and directly added to the simulated time-domain signals. Thermal noise in the circuit was represented by Johnson noise,

$$V_{Johnson} = \sqrt{4k_B T \Delta f R}, \quad (5)$$

where k_B is the Boltzmann constant, T is the absolute temperature, Δf is the bandwidth, and R is the circuit resistance. Here, R represents the AC resistance R_{AC} to account for the skin effect. This resistance increases with the resonance frequency ω_0 . An operating temperature of 300 K was assumed, and Δf and R were determined from the specific RLC configuration.

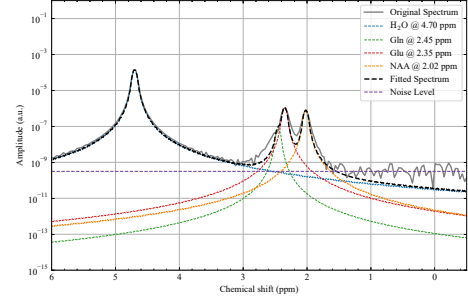


FIG. 2: Representative ¹H-MRS spectrum simulated at 0.3 T using a conventional RF coil. Despite the limited spectral resolution inherent to low-field conditions, the water peak and several major metabolite signals remain discernible.

OPM noise was considered to consist of optical and system components. Based on previous experimental configurations^{20,21}, the system noise was assumed to dominate the fundamental optical noise. Thus, the overall OPM noise was modeled as additive white Gaussian noise with an amplitude spectral density of $14 \text{ fT}/\sqrt{\text{Hz}}$.

A quantitative comparison between the proposed OPM-based system and a conventional induction-coil system was conducted in terms of SNR. The OPM output was simulated as a 2.5 s time-series signal, and fast Fourier transform analysis with a Hanning window was applied. SNR was defined as the peak amplitude of each metabolite signal divided by the baseline level, which comprised residual water, overlapping metabolite contributions, and background noise at the same spectral position. To examine field dependence, the static magnetic field strength of the MRI system was varied from 0.01 T to 7.0 T, and the SNR was calculated under each condition.

Figure 2 shows a representative ¹H-MRS spectrum of mixed metabolites calculated at 0.3 T using a conventional induction coil with IR. Despite the limited spectral resolution at low field, metabolite peaks are distinguishable from the dominant water peak. This demonstrates the feasibility of detecting major metabolites under practical low-field conditions, providing a baseline for subsequent simulation analyses.

The relative performance of IR and CHES water suppression methods was evaluated by comparing the SNR of Glu across different magnetic field strengths. Figure 3 shows that IR achieved higher SNR than CHES at fields below approximately 1 T. While CHES provided moderate suppression at high fields, its performance declined at low fields. In contrast, IR maintained better water suppression and enabled Glu detection down to 0.1 T. Furthermore, the magnetic field at which the dominant noise source transitioned from system noise to residual water signal was approximately 0.6 T for CHES and 0.1 T for IR. This indicates that IR achieves more effective water suppression under low-field conditions.

The field-dependent SNR of NAA was evaluated using

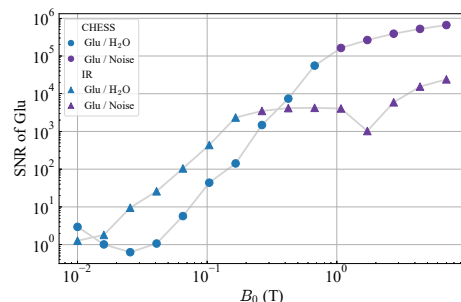


FIG. 3: Magnetic field dependence of Glu detectability under CHESS and IR. SNR values represent both Glu-to-water and Glu-to-noise ratios. IR maintains superior suppression efficiency below 1 T, enabling Glu detection down to 0.1 T (where IR-induced co-suppression occurs).

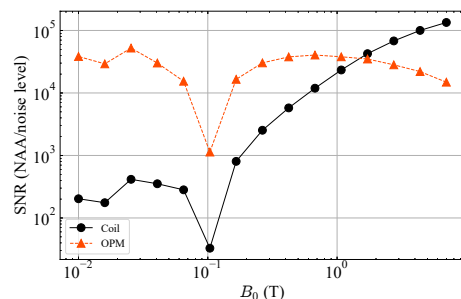


FIG. 4: Simulated SNR of NAA as a function of magnetic field strength. The results reveal a crossover point at 1 T, below which OPM outperforms coils. A distinct SNR dip at 0.1 T aligns with the convergence of T_1 relaxation times, suggesting a limitation of the IR method at this field strength.

single-metabolite simulations with both RF coil and OPM detection. The coil exhibited a power-law increase with field strength, while the OPM maintained nearly constant SNR across the examined range. Below 1 T, the OPM outperformed the coil. A distinct dip was observed at approximately 0.1 T for both detectors (Fig. 4), where the T_1 relaxation times of water and NAA converge.

The effects of detection scheme were explored by analyzing the signal amplitudes and the dominant noise sources of Gln, Glu, and NAA under mixed-metabolite conditions between OPM and coil detection (Fig. 5). As shown in the top panels, the OPM signal amplitude remained nearly constant in the low-field region, whereas the coil signal decreased significantly. Up to approximately 1 T, the system noise level remained the dominant limiting factor (bottom panels), and OPM detection demonstrated a clear advantage over RF coils detection in terms of the resulting SNR.

However, below 1 T, spectral overlap became increasingly prominent, causing the residual signals from other metabolites to dominate the noise floor and reducing the detectability of all metabolites. Adopting the standard 3σ criterion ($\text{SNR} \geq 3$) for reliable detection, the SNR of Glu dropped below 3 at 0.3 T, and those of Gln and NAA at 0.1 T, defining the practical lower bounds for quantification.

The observed performance difference underscores the limitations of frequency-selective techniques like CHESS below 1 T, highlighting the advantage of alternative contrast mechanisms such as IR. At low fields, reduced chemical shift separation necessitates longer, narrower-bandwidth RF pulses for CHESS, exacerbating relaxation-induced signal loss and reducing suppression efficiency. In contrast, the IR method is based on T_1 relaxation contrast, which is independent of frequency offset and thus maintains stable suppression performance even under low-field conditions. The IR method also suppresses signals from metabolites whose T_1 values are close to that of water.

The sharp dips observed in the SNR of certain metabolites are likely caused by the magnetic field strength at which T_1 relaxation time of the metabolite converges with that of water, resulting in co-suppression. To overcome this limitation, it is necessary to explore suppression methods that do not rely solely on T_1 values. For example, methods that utilize phase differences, such as the phase-sensitive Dixon technique²⁹, should be evaluated to identify suppression strategies better suited to low-field conditions.

The results confirm that OPM detection provides improved SNR compared with conventional RF coils in sub-tesla fields. This advantage in sensitivity becomes more pronounced as the static magnetic field strength decreases. This makes OPMs particularly suitable for low-field MRS systems targeting cost-effective metabolic diagnostics. This trend is theoretically consistent with the fundamental detection mechanisms of each detector. In the FT-OPM configuration, the NMR signal is first detected inductively by a pickup coil and therefore follows Faraday's law of induction. Consequently, the signal amplitude decreases with decreasing Larmor frequency, similarly to conventional RF coils.

However, the magnetic flux collected by the pickup coil is transferred through the flux transformer and detected by the OPM. As a result, the system noise is primarily determined by the OPM sensitivity rather than coil thermal noise, leading to a weaker SNR degradation at low magnetic fields. Additionally, the magnetic field threshold at which OPMs begin to outperform coils aligns with previous experimental studies¹⁹.

The present results revealed detection limitations due to spectral overlap, with SNR falling below 3 for Glu at 0.3 T and for Gln and NAA at 0.1 T. These findings highlight the necessity of improving spectral resolution and developing advanced separation techniques to ensure reliable metabolite quantification under low-field conditions. Since the SNR advantage of OPMs relative to RF coils becomes more pronounced with decreasing field strength, the development of techniques to minimize spectral interference from overlapping metabolites could further enhance the diagnostic utility of low-field OPM-MRS.

The current simulation treats major resonances as singlets

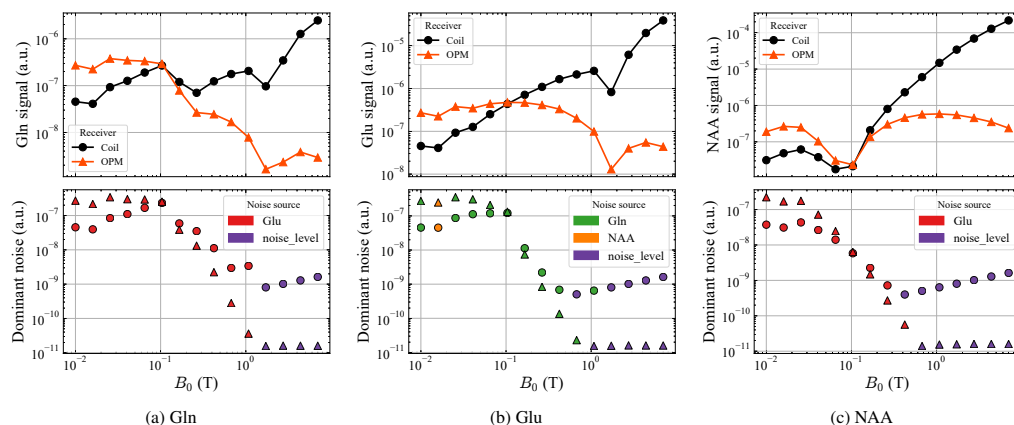


FIG. 5: Field-dependent behavior of signal and noise components for (a) Gln, (b) Glu, and (c) NAA in mixed-metabolite simulations. Each subfigure shows the metabolite signal amplitude (top) for coil and OPM detection, and the dominant noise components (bottom), distinguishing between system noise and residual metabolite noise. In all panels, marker shapes indicate the receiver type (circles for the RF coil and triangles for the OPM). In the bottom panels, marker colors represent the corresponding dominant noise sources. As the B_0 decreases, the signal amplitude of OPM detection remains nearly constant while the coil signal follows a power-law increase. Below 1 T, residual metabolite noise becomes dominant over system noise, leading to sharp reductions in SNR and detectability in the sub-tesla range.

to focus on fundamental detection sensitivity. It neglects multiplet splitting due to J-coupling and T_2^* -driven macroscopic line broadening. Consequently, the simulated spectral resolution at high magnetic fields may be overestimated compared to realistic *in vivo* environments. Efforts in future work should focus on evaluating quantification performance under these realistic conditions, enhancing frequency resolution, and validating the proposed methods through practical measurements using physical phantoms or prototype systems.

This study evaluated the feasibility of low-field ^1H -MRS using Bloch equation-based simulations. We demonstrated that OPM-based detection outperformed conventional RF coils in sub-tesla magnetic fields, maintaining stable SNR across 0.01–1 T. The IR method, based on T_1 contrast rather than frequency selectivity, enabled effective water suppression down to 0.1 T and enhanced Glu detectability below 0.7 T. These results suggest that OPM-MRS systems can support portable, low-cost implementations for metabolic detection in resource-limited settings.

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¹J. A. Frahm, H. Bruhn, M. Gyngell, K. Merboldt, W. Hänicke, and R. Sauter, *Magnetic Resonance in Medicine* **9**, 79 (1989).

- ²V. Mlynárik, M. Cacquevel, L. Sun-Reimer, S. Janssens, C. Cudalbu, H. Lei, B. L. Schneider, P. Aebischer, and R. Gruetter, *Journal of Alzheimers Disease* **31**, S87 (2012).
- ³A. Marsman, R. C. Mandl, D. W. Klomp, M. M. Bohlken, V. O. Boer, A. Andreychenko, W. Cahn, R. S. Kahn, P. R. Luijten, and H. E. H. Pol, *NeuroImage: Clinical* **6**, 398 (2014).
- ⁴G. Delvecchio, J. Stanley, A. Altamura, and P. Brambilla, *Epidemiology and Psychiatric Sciences* **26**, 587 (2017).
- ⁵C. Berthet, H. Lei, J. Thevenet, R. Gruetter, P. J. Magistretti, and L. Hirt, *Journal of Cerebral Blood Flow & Metabolism* **29**, 1780 (2009).
- ⁶World Health Organization, *Global Atlas of Medical Devices 2022* (World Health Organization, Geneva, 2022) licence: CC BY-NC-SA 3.0 IGO.
- ⁷M. Jalloul, M. Miranda-Schaubinger, A. M. Noor, J. M. Stein, R. Amiruddin, H. M. Derbew, V. L. Mango, A. Akinola, K. Hart, J. Weygand, *et al.*, *NMR in Biomedicine* **36**, e5022 (2023).
- ⁸J. P. Marques, F. F. Simonis, and A. G. Webb, *Journal of Magnetic Resonance Imaging* **49**, 1528 (2019).
- ⁹M. Hori, A. Hagiwara, M. Goto, A. Wada, and S. Aoki, *Investigative radiology* **56**, 669 (2021).
- ¹⁰T. C. Arnold, C. W. Freeman, B. Litt, and J. M. Stein, *Journal of Magnetic Resonance Imaging* **57**, 25 (2023).
- ¹¹D. Belkić and K. Belkić, *Journal of Mathematical Chemistry* **62**, 1251 (2024).
- ¹²A. Haase, J. Frahm, D. Matthaei, W. Hänicke, and K.-D. Merboldt, *Journal of Magnetic Resonance* **67**, 258 (1985).
- ¹³I. Tkáč, Z. Starčuk, I.-Y. Choi, and R. Gruetter, *Magnetic Resonance in Medicine: An Official Journal of the International Society for Magnetic Resonance in Medicine* **41**, 649 (1999).
- ¹⁴T. B. Moraes, F. V. C. Kock, K. S. Salome, A. Barison, A. Simpson, and L. A. Colnago, *Journal of Magnetic Resonance Open* **14**, 100090 (2023).
- ¹⁵M. Saracanie and N. Salameh, *Frontiers in Physics* **8**, 172 (2020).
- ¹⁶G. M. Bydder and I. R. Young, *J Comput Assist Tomogr* **9**, 659 (1985).
- ¹⁷J. Allred, R. Lyman, T. Kornack, and M. V. Romalis, *Physical Review Letters* **89**, 130801 (2002).
- ¹⁸I. K. Kominis, T. W. Kornack, J. C. Allred, and M. V. Romalis, *Nature* **422**, 596 (2003).

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- ¹⁹I. Savukov, S. Seltzer, and M. Romalis, *Journal of Magnetic Resonance* **185**, 214 (2007).
- ²⁰I. Hilschenz, Y. Ito, H. Natsukawa, T. Oida, T. Yamamoto, and T. Kobayashi, *Journal of Magnetic Resonance* **274**, 89 (2017).
- ²¹I. Savukov and T. Karaulanov, *Journal of Magnetic Resonance* **231**, 39 (2013).
- ²²F. Bloch, *Physical Review* **70**, 460 (1946).
- ²³J. R. Moffett, B. Ross, P. Arun, C. N. Madhavarao, and A. M. Nambodiri, *Progress in neurobiology* **81**, 89 (2007).
- ²⁴M. Warepam, K. Ahmad, S. Rahman, H. Rahaman, K. Kumari, and L. R. Singh, *Biomolecules* **10**, 286 (2020).
- ²⁵S. Elsaid, D. S. Rubin-Kahana, S. Kloiber, S. H. Kennedy, S. Chavez, and B. Le Foll, *International Journal of Molecular Sciences* **23**, 4754 (2022).
- ²⁶N. Bloembergen, E. M. Purcell, and R. V. Pound, *Physical Review* **73**, 679 (1948).
- ²⁷A. L. Lopez-Kolkovsky, S. Mériaux, and F. Boumezeur, *Magnetic Resonance in Medicine* **75**, 503 (2016).
- ²⁸L. An, S. Li, and J. Shen, *Magnetic Resonance in Medicine* **78**, 2073 (2017).
- ²⁹W. T. Dixon, *Radiology* **153**, 189 (1984).