Nuclear magnetic resonance and its use in magnetic resonance imaging (MRI)

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Ecole polytechnique fédérale de Lausanne - Travaux pratiques III, May 2024

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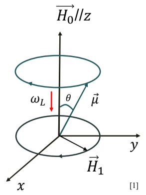
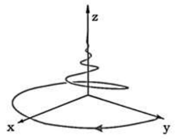
Motivation

- Nuclear Magnetic Resonance allows to probe matter using nuclear spin
- It can be used for non-invasive imaging (MRI)

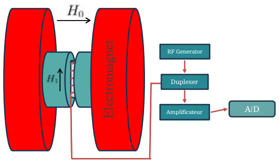


- The nuclear spin precess at Larmor frequency $\omega_L = \gamma_I H_0$, γ_I being the gyromagnetic ratio
- When an oscillating field $H_1 \ll H_0$ is applied perpendicularly, the spin rotates.
- The duration of H_1 determines the angle of rotation.

- Static field H_0 generated with an electromagnet
- H_1 field created by a coil
- Same coil used to measure signal



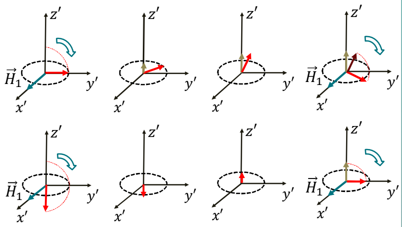
Magnetisation's dynamic of the return to equilibrium [1]

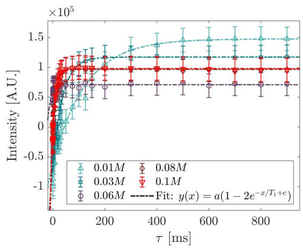


- 2 types of spin interactions cause a relaxation of the magnetization :
  - Spin-lattice : $M_z(t) = M_0 + (M_z(0) - M_0)e^{-t/T_1}$
  - Spin-spin : $\|\overrightarrow{M_{xy}}(t)\| = \|\overrightarrow{M_{xy}}(0)\|e^{-t/T_2}$
- $\overrightarrow{M_z}, \overrightarrow{M_{xy}}$ are to the projection of the magnetization in the corresponding direction/plane, M_0 is the equilibrium magnetization.

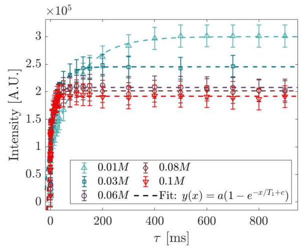
T1 measurement – Inversion & Saturation Recovery

- **Saturation recovery** sequence consists of 90° pulse and a measurement after a delay τ of the regenerated M_z component. 90° pulse is applied, the signal is recorded. Varying τ allows to recover the relaxation time T_1.
- **Inversion recovery** sequence replaces the initial pulse of the saturation recovery with a 180° pulse
- Most used sequences to measure the spin-lattice relaxation time T_1.





Magnetization of samples with varying concetrations using the **inversion recovery sequence** as a function of τ, with fit determining T_1 .



Magnetization of samples with varying concetrations using the **saturation recovery sequence** as a function of τ, with fit determining T_1 .

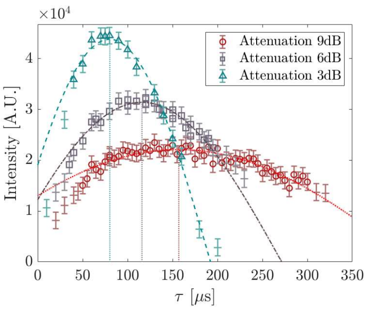
Sample	0.01M	0.03M	0.06M	0.08M	0.1M
T_1^{sat}[ms]	100 ±30	35±4	17±5	13.9±1.6	10.6±1.5
T_1^{inv}[ms]	120±20	38±11	13±13	13±2	12.3±0.6

Calibration of the 90° pulse

The time interval of the H_1 field that results in a 90° rotation of the magnetization is sought. It is determined by finding the interval leading to a **maximal response signal** (the projection of the magnetization on the plane orthogonal to H_0 is measured with the signal.)

- 5 samples of copper sulfate ($CuSO_4$-$5H_2O$) of different concentrations are created.
- Calibration : measure the intensity of the resulting signal for each samples as a function of τ, using different attenuations (dB).
- Sinusoidal fit applied, maximum recovered

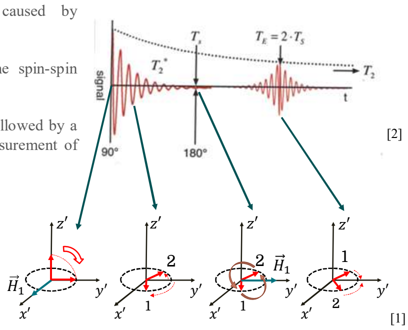
Attenuation	3dB	6dB	9dB
Pulse length [μs]	80±3	115.9±1.0	156.8±1.9

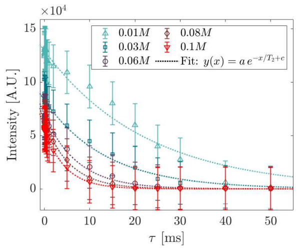


Intensity of transverse magnetization as a function of τ for different attenuations, with fit determining the pulse length.

T2 measurement – Spin-Echo sequence

- The observed decay time T_2^* is considerably shorter than T_2. This is due to the spin phase shift caused by inhomogeneities of the H_0 field.
- To isolate the T_2 relaxation time due to the spin-spin interactions alone, spin-echo sequence is used.
- **Spin-echo sequence** consists of a 90° pulse followed by a 180° pulse, separated by a delay T_S. The measurement of the echo is performed at the time $2T_S$.



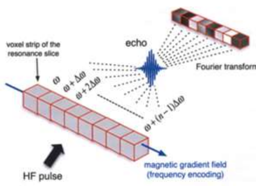


Magnetization of samples with varying concetrations using the **spin echo sequence** as a function of τ, with fit determining T_2 .

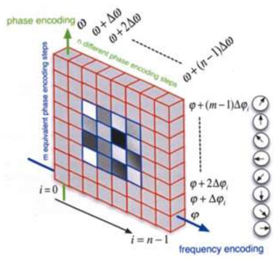
Sample	0.01M	0.03M	0.06M	0.08M	0.1M
T_2[ms]	23±3	14±2	7.2±0.7	5.6±0.5	4.6±0.3

Magnetic resonance imaging (MRI)

- Nuclear magnetic resonance is the basis of MRI.
- This technique unveils different materials in a sample by taking advantage of their different relaxation times T_1 and T_2.
- A magnetic field gradient is added to encode the position in one direction of the sample. This gradient lead to position-dependent Larmor frequencies. The position is then recovered with a Fourier transform of the measured signal. This technique is **frequency encoding**

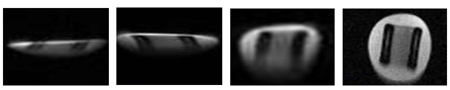


Frequency encoding for 1D imaging [2]

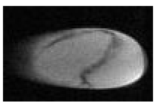


Frequency encoding and phase encoding for 2D imaging [2]

- To produce a 2D image, frequency encoding alone is not sufficient : applying two linear magnetic gradient fields in two directions would lead to a non-invertible signal (multiple points in space would share the same Larmor frequency).
- **Phase encoding** : a brief linear magnetic gradient is applied in the second direction to induce a dephasing. The position is recovered with a 2D Fourier transform.

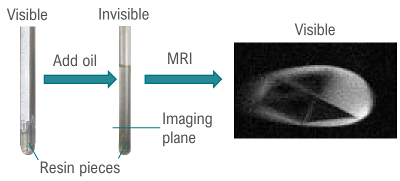


Results of the imaging of a teflon piece with varying phase encoding resolution. The resolution affect the number of vertical pixels, and thus the stretch of the final image



Organic tissue can be detected

- Resin pieces & oil have similar refraction index
- Submerged in oil it is not visible
- MRI allows to see the pieces in oil



Conclusions

- Calibration of H_1 duration
- Relaxation time T_1 measured using Inversion recovery and Saturation recovery, similar result obtained
- Relaxation time T_2 measured using spin echo
- $T_2 < T_1$ as expected
- Relaxation rate depends on concentration of copper sulphate (paramagnetic) : contributes to magnetisation
- Exploration of RMN use in the MRI process
- NMR replaces X-ray on molecular structure measurement [3], used for thermometer [4]

References

[1] EPFL, *Résonance magnétique nucléaire «Théorie et manuel pratique»*

[2] PHYWE, Laboratory Experiments, *Magnetic Resonance Imaging (MRI) I*

[3] Yizhou Liu et al. , *Unequivocal determination of complex molecular structures using anisotropic NMR measurements*. Science356, eaam5349 (2017)

[4] Ammann et al. , *A Simple Multinuclear NMR Thermometer* , Journal of Magnetic Resonance 46, 319-321 (1982)