

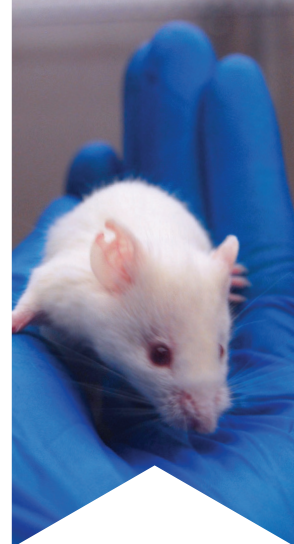
FFC APPLICATION NOTE: IN VIVO

NMRD as a biomarker for tumours: a preclinical study

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Introduction

Fast field cycling (FFC) NMR relaxometry is a low-field magnetic resonance technique which measures the dependence of the spin-lattice relaxation rate $R_1 (= 1/T_1)$ on the magnetic field over a wide range of field strengths with just one instrument. The important information extracted from the relaxation dispersion curves (NMRD profiles) concerns molecular motions (molecular dynamics) described by means of spectral density $J(\omega) \propto R_1$.

In vivo application of FFC NMR to the study of tumours

Fast Field Cycling NMR relaxometry is a versatile technique that can be applied to a wide number of sectors including the biomedical sector. In the last few years, several publications about promising FFC studies performed on *ex-vivo* samples have been published [1, 2, 3, 4]. However, in 2018 the first *in-vivo* FFC study performed on tumour-bearing mice was successfully carried out showing encouraging results, [5] with the possibility to discriminate between not only healthy and disease tissues but also between different kinds of tumours.

In vivo FFC reveals the role of intracellular water lifetime as a tumor biomarker

In [5] it is shown that there is a clear correlation between the rate of cellular water exchange and tumour activity. Furthermore, results show that there is also a correlation between relaxation rates and tumour types. In [5] a non-invasive technique based on magnetic resonance imaging (MRI) is presented which, using

weighted proton resonance dispersion profiles at low fields, allows tumour development to be monitored.

The *in-vivo* experiment

NMRD profiles were acquired on narcotized mice using a Stelar SPINMASTER FFC NMR relaxometer with a prototype set-up which allowed it to host a mouse (ca. 20gr). A 0.5T wide bore magnet was used with the implementation of a dedicated 11mm transmitter/receiver solenoid detection coil placed around the mouse's leg where the tumour cells were injected. In this study, 3 different kinds of mouse mammary adenocarcinoma cells were used: TS/A, 4T1, 168FARN. The 3 cell lines were selected because they have different aggressiveness and metastatic potential (i.e. 168FARN < TS/A < 4T1). The size of the tumour was comparable to the size of the leg, in order to be sure that the NMR signal from inside the coil was from the tumour and not from the healthy tissue. During the acquisition, the values of R_1 for 25 field points logarithmically distributed in the range [0.01-20MHz] were measured.

FIG. 1:
Anesthetized mouse in the probe, ready to be inserted inside the magnet for measurements. The white arrow indicates the detection ring coil around the tumour-bearing leg.

FIG. 2:
Anesthetized mouse in the probe, ready to be inserted inside the magnet for measurements. The white arrow indicates the detection ring coil around the tumour-bearing leg.



FIG. 1

FIG. 2

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The model

A tissue or an in-vitro cellular system may be considered as a two-compartment system, composed of:

- the **extracellular space** (comprising the intravascular volume) characterized by an averaged $R1_{ex}$ value
- the **intracellular compartment** characterized by a more restricted mobility of the water molecules, with relaxation termed $R1_{in}$
- V_{ex} and V_{in} being the respective volume fractions

Water molecules can cross the barrier between the 2 compartments mixing their relaxation rates. Therefore, the t_{in} (intracellular water residence time) and t_{ex} (extracellular water residence time) have to be inserted in the model. According to this bicompartamental model, the evolution time of M_z (longitudinal magnetization) is dependent on the relationship between the absolute values of the “relaxation” term, $|R1_{in} - R1_{ex}|$, and an “exchange” term $|k_{in} + k_{ex}|$, (where $K_{in} = 1/t_{in}$ and $k_{ex} = 1/t_{ex}$) in a relationship that has been previously defined as the NMR “shutter-speed”. In the intermediate-exchange region, the time evolution of M_z can be bi-exponential, but the relaxation rates obtained from the fitting of the data can be “contaminated” by the exchange occurring between the two compartments. Results showed that different intracellular water residence time t_{in} and intra- and extra-cellular volumes were obtained from the fitting procedure, with values that change proportionally to the tumour aggressivity.

	V_{ex}	t_{ex}
MUSCLE LEG	0.14±0.02	1.24±0.25
4T1	0.22±0.08	0.68±0.20
TS/A	0.20±0.02	0.99±0.19
168FARN	0.15±0.01	1.12±0.32

TAB. 1:
Values of extracellular volume fraction (V_{ex}) and intracellular water residence time for healthy tissue and 3 different types of tumours. (Adapted from [5]).

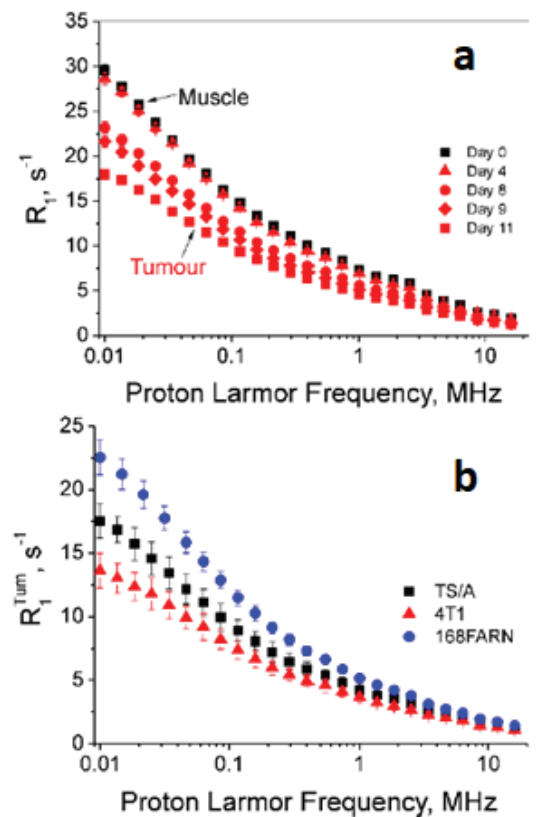
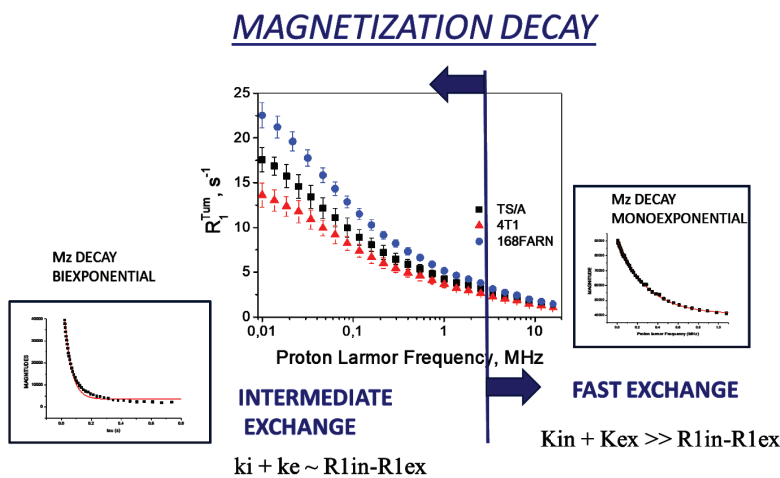


FIG. 4:
a) NMRD profile of a mouse leg tissue: before (day 0) and after the intramuscular injection of 1 million 4T1 cells.
b) NMRD profiles of the tumour tissues grown on hind-limbs: 4T1 (red filled triangle), TS/A (black filled circle) and 168FARN (blue filled circle) acquired 11 ± 2 , 13 ± 3 and 25 ± 1 days after intramuscular injection, respectively. $R1^{Tum}$ is the averaged relaxation rate normalized to the tumour mass fraction compared to the whole hind-limb (Adapted from [5]).



The magnetization recovery depends on :
 $R1_{IN}, R1_{EX}, V_{IN}, V_{EX}, k_{IN}, k_{EX}$

Conclusions

The following important points are worth mentioning:

1. The field dependence of tissue water proton T_1 at low magnetic fields reveals new diagnostic information.
2. The intracellular water lifetime acts as a molecular biomarker reporting on cellular metabolism and expression of membrane transporters.
3. The limitation of the absence of spatial resolution will be partially overcome using a surface coil (in collaboration with Stelar) or using the FFC-MRI scanner built by D. Lurie and L. Broche [6].
4. Decrease of T_1 indicates that the water exchange rate across the membrane is a distinctive hallmark that differentiates muscle (representative of healthy cells) and tumour cells.



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