Quantitative proton MRI and MRS of the rat brain with a 3 T clinical MR scanner

M. Aradia,a,b,∗, R. Steiera,b, P. Bukovicsa,b, C. Szalayc, G. Perlakib, G. Orsib, J. Pál,a, J. Janszkyd, T. Dóczia, A. Schwarcza,a,b

a Department of Neurosurgery, University of Pécs, Rét u. 2, 7624 Pécs, Hungary
b Pécs Diagnostic Institute, University of Pécs, 7624 Pécs, Hungary
c Institute of Physiology, University of Pécs, 7624 Pécs, Hungary
d Department of Neurology, University of Pécs, 7624 Pécs, Hungary

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Rat; Brain; MRI; Spectroscopy; Clinical scanner

Summary
Objective. — To demonstrate the capability of a clinical 3 T human scanner in performing quantitative MR experiments in the rat brain.

Material and methods. — In vivo, measurements on eight Wistar rats were performed. Longitudinal relaxation time (T1) and transverse relaxation time (T2) measurements were set up at a spatial resolution of 0.3 × 0.3 × 1 mm3. Diffusion-weighted imaging was also applied and the evaluation included both mono- and biexponential approaches (b-value up to 6000 s/mm2). Besides quantitative imaging, the rat brain was also scanned at a microscopic resolution of 130 × 130 × 130 μm3. Quantitative proton spectroscopy was also carried out on the rat brain with water as internal reference.

Results. — T1 and T2 for the rat brain cortex were 1272 ± 85 ms and 75 ± 2 ms, respectively. Diffusion-weighted imaging yielded accurate diffusion coefficient measurements at both low and high b-value ranges. The concentrations of MR visible metabolites were determined for the major resonances (i.e., N-acetyl-aspartate, choline and creatine) with acceptable accuracy.

Conclusion. — The results suggest that quantitative imaging and spectroscopy can be carried out on small animals on high-field clinical scanners.

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Introduction

Magnetic resonance imaging (MRI) in animals has the substantial advantage that the evolution of a modeled pathology can be followed in vivo. The complicated signal
changes detected in human pathologies (e.g., multieponential diffusion) can be modeled and the background of these signal changes may be elucidated via the morphology [1]. However, the dedicated small-animal (i.e., rat and mouse) scanners with strong gradient coils can be quite expensive. Many clinically-based researchers, and especially radiologists, do not have access to such MR apparatus. To circumvent this situation much effort has been devoted proving that large-bore clinical MR scanners have the ability to perform small-animal imaging with acceptable quality [2—17]. Most of the reported successful rat MRI measurements performed with human scanners provided merely qualitative data: only $T_1$ or $T_2$-weighted images were produced [9] together with quantitative ADC data [6,18]. The reason for refraining from quantitative MRI is that the low signal-to-noise ratio (SNR) and inappropriate MR sequences may yield very inaccurate results. This is also obvious in the case of dedicated small-animal scanners operating at higher magnetic field strength, e.g., the reported $T_1$ values for the rat brain range from 928 ms to 1690 ms at 4.7 T in the literature [19,20]. This situation demonstrates the merit of a recently published paper in which $T_1$, $T_2$ and magnetization transfer were determined in vitro with high accuracy in different tissue types at 3 T [21]. These latter authors state that the literature regarding $T_1$ and $T_2$ data at 3 T is surprisingly limited. Quantitative MRI would be important, for instance, in brain edema studies in which the extent of the edema appears identical (i.e., the same hyperintense area is seen on the images) but $T_1$ and $T_2$ may be significantly different among animals depending on the water content [22,23]. As regards the spectroscopy of small animals on human scanners, only a few papers are available in the literature [24,25].

Combination of the relatively high SNR of a 3 T human scanner with a strong insert gradient coil make DTI feasible in the rat brain [26]. It was demonstrated that a field strength of 3 T allows not only better human imaging, but also small-animal imaging with a spatial resolution comparable to those obtained with dedicated small-bore magnets [9,12,27]. Accordingly, 3 T clinical scanners may readily be employed in small-animal imaging, and expensive small-bore systems are not necessary for most applications. The advantage of a clinical MRI scanner is that it has very homogenous $B_1$ field; however, the smaller gradient amplitude and the lack of flexibility in pulse sequences imposes problems to overcome.

The aim of our study was to demonstrate the capability of a 3 T human scanner in performing quantitative imaging in small-animals. The study was carried out using commercial coils and commercial MR sequences; no sequence manipulation or extra MR hardware was applied.

Methods

Phantom experiments

A group of solutions was prepared with increasing gadolinium (Magnevist, Bayer®) concentrations. The solutions were measured by three independent imaging methods in order to test the accuracy of the measurement of $T_1$: (i) saturation recovery spin echo imaging single slice, (ii) inversion recovery spin echo imaging single slice and (iii) multislice inversion recovery turbo spin echo imaging. Method (iii) was then used in an animal study (see below). The same coil setup was used for phantom and animal experiments (i.e., no major difference in filling factor).

Animals

The experiments were carried out on eight Wistar rats (250—300 g, three females, five males). The rats were intubated and anesthetized with 1.5% isoflurane in a 70/30 mixture of $N_2O$ and $O_2$. The rectal temperature of the animals was maintained between 36.5 and 37.5 °C throughout the experiments by using an isothermal pad (Deltaphase®, Braintree Scientific Inc., Braintree, USA). The head of each rat was fixed in a custom-built plastic head holder and placed through a loop RF coil (Fig. 1). After MRI and MRS examinations, the animals were sacrificed with an overdose of isoflurane. The brains were rapidly removed in a humid chamber, brains weighed immediately and dried to constant weight at 90 °C. The percentage water content ($W$) of each sample was calculated according to Eq. (1):

$$W = 100 \times \left( \frac{\text{wet weight} - \text{dry weight}}{\text{wet weight}} \right)$$

(1)

was utilized in the quantitative MRS calibration.

MRI and MRS

MRI was performed on a 3.0 T clinical MRI scanner (Magnetom TIM Trio, Siemens Medical Solutions, Erlangen, Germany) with a field gradient strength of 40 mT/m. Excitation was performed with the body coil, while a commercial loop RF coil with an inner diameter of 40 mm (Siemens Medical Solutions, Erlangen, Germany) was applied for signal detection. After the acquisition of scout images of the rat brain, $T_1$, $T_2$, and diffusion measurements were performed. The MRI protocol was followed by MRS experiments. High-resolution $T_2$-weighted MRI was also performed in three rats in separate experiments carried out by a turbo spin echo imaging sequence:

$$TR/TE = 3000/76 \text{ ms}, \hspace{1em} \text{slice thickness} = 0.13 \text{ mm}, \hspace{1em} FOV = 33 \times 33 \text{ mm}^2, \hspace{1em} 256 \times 256 \text{ pixel matrix, bandwidth} = \text{...}$$

Figure 1 The custom-built head holder with the 40 mm (i.d.) loop coil. The rat is intubated and anesthetized.
40 Hz/pixel, fat suppression with spectral attenuated inversion recovery method (SPAIR), number of acquisitions = 1.

$T_1$ was determined by an inversion recovery method, applying a turbo spin echo imaging sequence: slice selective inversion pulse, $TR/TE = 4200/15$ ms, slice thickness = 1 mm, distance factor = 0 mm with two concatenations (i.e., no gap, two interleaved series of slices), $FOV = 31 \times 31$ mm$^2$, $128 \times 128$ pixel matrix, bandwidth = 300 Hz/pixel, number of acquisitions = 1, echo train length = 8, number of slices = 20, inversion times = 300, 700, 1300, 1900, 2400 and 2800 ms. $T_2$ was determined by the standard spin echo imaging version of the CPMG sequence: $TR/TE = 2600/12.1$, $24.2$, $36.3$, $48.4$, $60.5$, $72.6$, $84.7$, $96.8$, $108.9$ and $121$ ms, slice-thickness = 1 mm, distance factor = 0 mm (i.e., no gap), $FOV = 33 \times 33$ mm$^2$, $128 \times 128$ pixel matrix, bandwidth = 175 Hz/pixel, number of acquisitions = 1, number of slices = 20.

Diffusion was determined with a trace-weighted single-shot echo planar imaging sequence: $TR/TE = 4000/119$ ms, slice thickness = 1.9 mm, distance factor = 0 mm (i.e., no gap), $FOV = 30 \times 95$ mm$^2$, $40 \times 128$ pixel matrix, bandwidth = 830 Hz/pixel, number of acquisitions = 4, number of slices = 14, $b$ values: 0, 500, 1000, 2000, 3000, 4000, 5000 and 6000 s/mm$^2$.

Before MRS acquisition, a 10 $\times$ 10 $\times$ 5 mm voxel was defined in the middle of the rat brain (Fig. 2). After localized manual shimming and water suppression adjustment, fully relaxed short-echo-time proton MR spectra (PRESS, $TR/TE = 6000/30$ ms, 64 accumulations) were acquired. Water suppression was accomplished with a chemical shift-selective sequence [CHESS] pulse. At the end of the MRS experiment, a reference water signal for the calibration of metabolite concentration was also acquired by turning off the water suppression. The total experimental protocol lasted for ~36 min comprising (i) 13 min for $T_1$ measurement, (ii) 4 min for $T_2$ measurement, (iii) 6 min for ADC quantification, (iv) 13 min for quantitative proton spectroscopy including manual adjustments. The acquisition time for high-resolution $T_2$-weighted imaging was 146 min.

**Data processing**

$T_1$ maps were calculated pixelwise from the images with different inversion times applying either a standard two-parameter fit (Eq. (2) [28]) or a three-parameter fit that accounts for RF pulse imperfections (Eq. (3) [29,30]).

$$M = |M_0 \times (1 - 2 \times \exp(-TI/T_1) + \exp(-TR/T_1))|$$

where $M$ is the measured signal intensity, $M_0$ is the signal intensity at thermal equilibrium, $TI$ is the inversion time, and $TR$ is the repetition time.

$$M = |A + B \times \exp(-TI/T_1)|$$

where $A$ and $B$ are constants incorporating $M_0$ and corrections for imperfect RF pulses (i.e., deviation from the nominal flip angle).

$T_2$ maps were produced from the pixelwise fit of Eq. (4), assuming monoexponential signal decay:

$$M = M_0 \times \exp(-TE/T_2))$$

ADC maps were also generated by a monoexponential fit of signal intensities in the low $b$ value range (i.e., $b$ value = 0, 500 and 1000 s/mm$^2$). A manual coregistration of ADC maps with $T_1$ maps was performed in each slice (i.e., two-dimensional stretching) in order to decrease the EPI distortion artifacts. The ADC values of the rat cortex were broken down into $ADC_{fast}$ and $ADC_{slow}$ by applying a biexponential fit throughout the whole $b$ value range (Eq. (5)).

$$M/M_0 = f_{fast} \exp(-b \times ADC_{fast}) + f_{slow} \exp(-b \times ADC_{slow})$$

where $M$ is the signal in the presence of diffusion sensitization, $M_0$ is that in the absence of diffusion sensitization, $ADC_{fast}$ and $ADC_{slow}$ are apparent diffusion coefficient values, and $f_{fast}$ and $f_{slow}$ are the contributions to the signal of the fast and slow-diffusing water compartments. Freehand regions of interests (ROIs) were drawn on $T_1$, $T_2$ and ADC maps to yield mean values (Fig. 3).

Raw spectroscopic data were postprocessed by using a Siemens Leonardo Workstation with built-in Siemens spectroscopy software. We applied a Hanning filter and zero filling to 2048 data points before Fourier transformation. After manual baseline and phase correction, an automatic fit was applied to the Fourier-transformed spectra, and the peak integrals of water, choline (Cho), creatine (Cr) and N-acetyl-aspartate (NAA) were determined.

The molar tissue water content ($MWC$) was calculated from the tissue water content ($W$) determined by the wet/dry method and the molar concentration of pure water (55.6 mol/L). Finally, the absolute metabolite tissue concentrations were calculated from

$$C = I_m \times MWC \times \frac{2}{(n \times I_W)}$$

where $C$ is the molar concentration of the metabolite, $n$ is the number of resonating protons, $I_m$ is the peak integral of a given metabolite, and $I_W$ is the peak integral of water. This formula only applies for spectra with long TR and short TE.
Figure 3  $T_1$ maps of the rat brain generated by two- (a) or three- (b) parameter fit of the turbo spin echo inversion recovery data. The $T_1$ map generated by the two-parameter fit demonstrates more homogeneity in the cortex, and accordingly a lower SD (see Table 2). The regions of interest (ROI) used for the cortex are shown (a). $T_2$ (c) and ADC (d) maps of the same slice are also presented.

Table 1  $T_1$ values (ms) obtained for gadolinium solutions (mmol/l). Turbo spin echo inversion recovery data were fitted by using either a two- (Fit1) or a three-parameter fit (Fit2). The equations for the fits are presented in the Methods section. Means and standard deviations are presented.

<table>
<thead>
<tr>
<th>Gadolinium concentration</th>
<th>Spin echo saturation recovery single slice</th>
<th>Spin echo inversion recovery single slice</th>
<th>Turbo spin echo inversion recovery multislice</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.266</td>
<td>915 ± 2</td>
<td>909 ± 3</td>
<td>909 ± 11</td>
</tr>
<tr>
<td>0.2</td>
<td>1184 ± 3</td>
<td>1133 ± 13</td>
<td>1078 ± 32</td>
</tr>
<tr>
<td>0.133</td>
<td>1440 ± 2</td>
<td>1335 ± 18</td>
<td>1360 ± 19</td>
</tr>
<tr>
<td>0.083</td>
<td>1865 ± 5</td>
<td>1815 ± 28</td>
<td>1726 ± 11</td>
</tr>
<tr>
<td>0.053</td>
<td>2038 ± 10</td>
<td>2145 ± 45</td>
<td>1997 ± 21</td>
</tr>
</tbody>
</table>

Statistical analysis

In phantom experiments only one ROI was used for each solution, standard deviation represents the variability of $T_1$ within the ROI. Concerning the animal study, two ROIs were placed in the cortex on both sides of the brain (Fig. 3). Within one animal, mean value and standard deviation was calculated by taking into account pixelwise data of both ROIs. The Coefficient of variation (CV) was calculated (i.e., ratio of the SD to the mean) to normalize the SD in animal experiments.

Results

Phantom experiments

All three methods, (i) saturation recovery spin echo imaging single slice, (ii) inversion recovery spin echo imaging single slice and (iii) multislice inversion recovery turbo spin echo imaging, yielded similar $T_1$ values with low standard deviation (Table 1). Both the two- and three-parameter fit gave accurate results in multislice inversion recovery turbo spin echo imaging (Table 1). Multislice inversion recovery turbo spin echo sequence was chosen for in vivo measurements because it provided a multislice acquisition in a reasonable time frame.

High-resolution MRI

Although the focus of our study was quantitative MRI and MRS, the high-resolution $T_2$ weighted image shows the capabilities of a 3T human scanner (Fig. 4). A microscopic resolution of $130 \times 130 \times 130 \mu m^3$ was achieved (Signal-to-noise ratio = 30).

$T_1$ and $T_2$ in the rat brain

$T_1$ maps (Fig. 3) were generated by applying either the two-parameter fit or the three-parameter fit. The two-parameter fit gave a mean $T_1$ value of 961 ± 2.5% ms for the cortex while the three-parameter fit gave significantly ($p < 4.3 \times 10^{-7}$, paired t-test) larger $T_1$ value of 1272 ± 6.7% ms (Table 2). The CV of $T_1$ was less in the two parameter
fit (Table 2). Fig. 5 demonstrates the difference between two- and the three-parameter fit in the in vivo data obtained from the cortex as free hand ROI. It is clear that the two-parameter fit is not a good model (\( r^2 = 0.96 \)), while the three-parameter fit gives a nearly perfect match (\( r^2 = 0.99 \)) (Fig. 5). However, due to the larger CV, the three-parameter fit resulted in less smooth \( T_1 \) maps than those produced via the two-parameter fit (Fig. 3).

The \( T_2 \) maps (Fig. 3) produced a mean \( T_2 \) value of 75 ± 2% ms for the rat cortex. The standard deviation was less than that in the \( T_1 \) measurement (Table 2).

### Diffusion

The ADC maps (Fig. 3) were produced by using only the low \( b \) value range (i.e., 0,500 and 1000 s/mm\(^2\)). The spatial resolution of ADC maps is much lower than that of \( T_1 \) and \( T_2 \) maps with obvious image blurring. The low image quality is mainly due to the gradient limitations of the scanner and not to the low SNR. The quantitative ADC data are presented in Table 2. The large voxel size in DWI produced a high SNR which allowed the acquisition of diffusion images with larger \( b \) values. The SNR was > 15 even on the images with \( b = 6000 \) s/mm\(^2\). Via the biexponential approach throughout the whole \( b \) value range \( ADC_{\text{fast}} \) and \( ADC_{\text{slow}} \) with the corresponding volume fractions were determined with acceptable standard deviation.

**Table 2** Quantitative MR parameters measured in the rat cortex. \( T_1 \) data were fitted using either two- (Fit1) or three- parameter fit (Fit2). The equations for the fits are presented in the Methods section. Means and coefficient of variations (CV) are presented.

<table>
<thead>
<tr>
<th></th>
<th>Fit1</th>
<th>Fit2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean ( T_1 )</td>
<td>961</td>
<td>1272</td>
</tr>
<tr>
<td>CV (% )</td>
<td>2.5</td>
<td>6.7</td>
</tr>
<tr>
<td>Mean ( T_2 )</td>
<td>75</td>
<td>6.02</td>
</tr>
<tr>
<td>CV (% )</td>
<td>2.0</td>
<td>3.8</td>
</tr>
<tr>
<td>( ADC_{\text{mono}} )</td>
<td>9.36</td>
<td>2.04</td>
</tr>
<tr>
<td>CV (% )</td>
<td>9.7</td>
<td>16</td>
</tr>
<tr>
<td>( ADC_{\text{fast}} )</td>
<td>74.3</td>
<td>6.1</td>
</tr>
<tr>
<td>CV (% )</td>
<td>25.7</td>
<td>17.5</td>
</tr>
<tr>
<td>( ADC_{\text{slow}} )</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Times \( T_1 \) and \( T_2 \) are shown in ms; \( ADC_{\text{mono}} \), \( ADC_{\text{fast}} \), \( ADC_{\text{slow}} \) units: (× 10\(^{-4}\) mm\(^2\)/s\(^{-1}\)); \( p_{\text{fast}} \) and \( p_{\text{slow}} \) units: %.

**Figure 5** Comparison of two- and three- parameter fits used for \( T_1 \) measurement. The equations for the fits are presented in the Methods section. The intensity (arbitrary units) obtained from a rat cortex is plotted against the inversion time (ms). Dashed and continuous lines correspond to the two- and three-parameter fits, respectively.

### Spectroscopy

The localized manual shimming to the voxel was essential in the signal acquisition. A representative spectrum of the rat brain is shown in Fig. 2. The SNR for the NAA peak was \( \sim 17 \) which could permit a smaller voxel size, but reduction of the voxel size was not possible because of limitations in the commercial PRESS sequence. The mean water concentration (\( W \)) in the rat brain was 78.0 ± 0.5%. It was used for calibration of the water peak in Eq. (6). The metabolite concentrations determined for NAA, Cr and Cho are presented in Table 3.

**Table 3** Metabolite concentrations (mmol/l) in the rat brain, measured by proton spectroscopy, means and coefficient of variations (CV) are presented.

<table>
<thead>
<tr>
<th></th>
<th>Cho</th>
<th>Cr</th>
<th>NAA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>1.77</td>
<td>6.25</td>
<td>8.78</td>
</tr>
<tr>
<td>CV (%)</td>
<td>21</td>
<td>25</td>
<td>14</td>
</tr>
</tbody>
</table>

### Discussion

The advances in the technology of clinical MR scanners permit high-quality small-animal imaging [2,12,26,27]. MR spectroscopy has also become possible in small animals in consequence of the higher magnetic field strength and custom-made coils [24,25]. It was recently demonstrated that mouse brain MRI is feasible with microscopic resolution even at the relatively low magnetic field of 2.35 T [31]. Although this was not the focus of our study, we succeeded in performing rat brain MRI with microscopic resolution. Long measurement time is usual in MRI anatomical studies of small animals [27,31].

Quantitative MRI and MRS data acquired with clinical scanners are seldom reported [32,33], though they would be important in the follow-up of in vivo pathologies. For instance, \( T_1 \) values of the brain may be converted to water content, so that brain edema can be measured in vivo [23,34–36]. However, the pitfall of in vivo \( T_1 \) measurements is clearly demonstrated by comparing the in vivo and in vitro data in the present study. Both two- and the three-parameter fits gave accurate \( T_1 \) values for the gadolinium solutions; however, the two-parameter fit underestimated \( T_1 \) in vivo. Fig. 6 shows the dependence of the published \( T_1 \) values for the rat cortex [19,20,33,35,37–51] on the magnetic field strength. Only empirical data are available for \( T_1 \) of brain tissue vs magnetic field [12]. A \( T_1 \) value of \( \sim 1270 \) ms for 3 T...
Quantitative rat brain MRI on a clinical scanner

Figure 6  This figure shows the dependence of the published $T_1$ values for the rat brain [19,20,33,35,37–50] on the magnetic field strength. There is a clear linear trend (continuous line) between $T_1$ and the magnetic field strength in the range of 1.5–11.4 T. The figure also shows the empirical relationship (dashed line) between $T_1$ and magnetic field strength derived from [51].

is estimated by fitting either a general equation of Rooney et al. [12] or a simple linear trendline (Fig. 6). When this predicted $T_1$ is taken into account, only the three-parameter fit furnishes accurate $T_1$ values. Failure of the two-parameter fit in vivo is probably explained by the RF imperfections which result in incomplete inversion and refocusing.

It should also be noted that the CV is larger when the three-parameter fit is used than that produced by the two-parameter fit. Nevertheless, this ∼7% CV obtained with a clinical scanner is still within the CV range observed for small-bore systems: 2–10% [40,42,43,52]. Further, on the basis of a meta-analysis of the human $T_1$ at 1.5 T, Tofts and Boulay [53] speculated that biological variation of $T_1$ in the brain should be around 5%.

Fig. 7 shows the literature $T_2$ values for the rat brain as a function of magnetic field strength [19,20,39–41,45,47–50]. $T_2$ data are often reported because of the direct relationship with brain edema [22] and also because of the simple and rapid measuring method. Similarly to $T_1$, the $T_2$ data also suggest a linear dependency on magnetic field strength (Fig. 6). This linear relationship indicates that $T_2$ for the rat brain should be ∼76 ms at 3 T which is in very good agreement with our data. The CV of $T_2$ is smaller (∼2%) than that of $T_1$, probably as a result of the lower sensitivity of measurement in RF field problems.

Quantitative, low $b$ value range $ADC$ data were earlier obtained on human scanners, mostly in stroke studies of the rat brain [6,18]. Although a good SNR can be achieved with the current set-up at 3 T, the limitations (mainly in gradients) did not allow better spatial resolution with the applied EPI sequence. Similar $ADC$ images with low spatial resolution obtained with clinical scanners have been reported in the literature at 3 T [12]. The high SNR originating from the low spatial resolution allows larger $b$ values. The biexponential analysis of the diffusion data yielded $ADC_{fast}$ and $ADC_{slow}$ values in very good agreement with those to be found in the literature [1,54,55]. $ADC_{mono}$, ($\times 10^{-4}$ mm² s⁻¹), $ADC_{slow}$ ($\times 10^{-4}$ mm² s⁻¹) and the volume fractions (%) for instance are almost identical to the results measured in a mouse study [55] at 9.4 T ($ADC_{mono}$: 6.02 ± 3.8% vs. 5.90 ± 3.4%; $ADC_{slow}$: 2.04 ± 16.0% vs. 2.02 ± 9.9%; $p_{fast}$: 74 ± 4.0% vs. 75 ± 4.0%).

It is intriguing that despite the markedly increased diffusion and echo time, the volume fractions do not change. Although there is a debate in the literature on the origin of the biexponential signal decay in diffusion measurements, our results clearly demonstrate a good reproducibility regardless of the field strength and echo time.

Besides the quantitative imaging experiments, we succeeded in performing quantitative proton MRS in the rat brain. The limitations in the Siemens spectroscopy sequence did not allow the smaller voxel size that could have been achieved according to the SNR of the metabolite spectra. The feasibility of MRS in the rat brain at a 3 T clinical scanner has already been demonstrated through the utilization of special, non-commercial hardware elements [24,25]. Although absolute quantification (i.e., in mmol/l) was not performed, metabolite concentrations were presented in institutional units, normalized to an arbitrary NAA value of 10.0 [44,45]. The metabolite concentrations obtained in our study are in overall agreement with earlier literature results [56,57]. Obviously, the postprocessing steps applied, (mostly baseline correction) and the quantification method (i.e., an external or an internal reference, the principle of reciprocity, etc.) might have a considerable impact on the absolute metabolite concentrations. Nevertheless, our metabolite concentrations and standard deviations almost perfectly match the values reported for the mouse brain using the same water reference strategy [58].

Conclusion

Quantitative $T_1$, $T_2$, $ADC$ (including a biexponential approach) and MRS data were obtained at 3 T on a human scanner. The measured $T_1$, $T_2$, $ADC$ data appear to be accurate and precise and also comparable to those obtained with dedicated small-bore systems operating at higher magnetic field strength. A microscopic resolution was achieved
in the rat brain as concerns structural imaging. Our results suggest that reliable small animal quantitative NMR measurements can be carried out on new-era high-field clinical scanners. The experiments do not require special MR hardware or software elements. However, the limitations of a clinical scanner in gradient strength and software flexibility should be always taken into account. The results will hopefully encourage others to carry out state-of-the-art rat MRI or MRS on clinical scanners.

Conflict of interest

None.

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