

# Magnetic resonance imaging T<sub>1</sub>- and T<sub>2</sub>-mapping to assess renal structure and function: a systematic review and statement paper

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## ABSTRACT

This systematic review, initiated by the European Cooperation in Science and Technology Action Magnetic Resonance Imaging Biomarkers for Chronic Kidney Disease (PARENCHIMA), focuses on potential clinical applications of magnetic resonance imaging in renal non-tumour disease using magnetic resonance relaxometry (MRR), specifically, the measurement of the independent quantitative magnetic resonance relaxation times T<sub>1</sub> and T<sub>2</sub> at 1.5 and 3 Tesla (T), respectively. Healthy subjects show a distinguishable cortico-medullary differentiation (CMD) in T<sub>1</sub> and a slight CMD in T<sub>2</sub>. Increased cortical T<sub>1</sub> values, that is, reduced T<sub>1</sub> CMD, were reported in acute allograft rejection (AAR) and diminished T<sub>1</sub> CMD in chronic allograft rejection. However, ambiguous findings were reported and AAR could not be sufficiently differentiated from acute tubular necrosis and cyclosporine nephrotoxicity. Despite this, one recent quantitative study showed in renal transplants a direct correlation between fibrosis and T<sub>1</sub> CMD. Additionally, various renal diseases, including renal transplants, showed a moderate to strong correlation between T<sub>1</sub> CMD and renal function. Recent T<sub>2</sub> studies observed increased values in renal transplants compared with healthy subjects and in early-stage autosomal dominant polycystic kidney disease (ADPKD), which could improve diagnosis and progression assessment compared with total kidney volume alone in early-stage ADPKD. Renal MRR is suggested to be sensitive to

renal perfusion, ischaemia/oxygenation, oedema, fibrosis, hydration and comorbidities, which reduce specificity. Due to the lack of standardization in patient preparation, acquisition protocols and adequate patient selection, no widely accepted reference values are currently available. Therefore this review encourages efforts to optimize and standardize (multi-parametric) protocols to increase specificity and to tap the full potential of renal MRR in future research.

**Keywords:** magnetic resonance imaging, kidney, mapping, relaxometry, chronic kidney disease

## INTRODUCTION

Kidneys are morphologically complex organs. Renal pathologies induce (micro-) structural and functional changes that may be captured with magnetic resonance imaging (MRI) owing to its exceptional soft tissue contrast. Despite the frequent and successful use of magnetic resonance relaxometry (MRR) in other organs (e.g. cardiac MRI) to assess oedema, amyloid deposition and fibrosis, the application of renal MRR is still scarce.

Renal MRR holds the promise to non-invasively quantify tissue inflammation and alterations, such as interstitial or cellular oedema and/or fibrosis, as well as renal function. This review article evaluates and summarizes data on renal T<sub>1</sub> and T<sub>2</sub> mapping using clinical 1.5 and 3Tesla (T) systems and provides

recommendations for upcoming research efforts to promote MRR in clinical practice.

## MATERIALS AND METHODS

The European Cooperation in Science and Technology (COST) Action Magnetic Resonance Imaging Biomarkers for Chronic Kidney Disease (PARENCHIMA) ([www.renalmri.org](http://www.renalmri.org)) initiated this systematic review by an extended PubMed search regarding renal mapping (see [Supplementary data](#)) on 25 October 2017 to identify human *in vivo*  $T_1$  and  $T_2$  measurements at 1.5 and 3T. Titles and abstracts of 357 publications were processed to identify matches aligning with the aim of this article. Furthermore, relevant references within the acquired papers and selected studies by the authors were added. Our analysis reaches back to the year 1983 and includes studies with field strengths below 1.5T. Some handpicked qualitative studies and preclinical studies were also included to present readers with relevant trends in the measurement of renal  $T_1$  and  $T_2$  values. Studies regarding renal neoplasms and/or dynamic contrast-enhanced MRI were excluded. For details on data collection, see [Supplementary data](#).

## BASIC PRINCIPLES OF MAGNETIC RELAXATION MECHANISMS

MRI is a non-invasive technique to map the human body using the interaction of three magnetic fields: (i) a strong static field ( $B_0$  or main magnet) to magnetize the whole sample and to allow the signal to be measured; (ii) gradient coils producing three ( $G_x$ ,  $y$ ,  $z$ ) linear, orthogonal gradients to allow the signal to be registered in space; and (iii) a dynamic radio frequency (RF) field ( $B_1$  or excitation field) to change steady-state magnetization produced by  $B_0$  and to enable the readout of the measured signal (using an appropriately frequency-tuned coil or antenna) [1].

When subjects are placed inside the MRI scanner, nuclear spins align with  $B_0$  ([Figure 1a and b](#)). The application of an RF pulse ( $B_1$ ; usually in the range of milliseconds and millitesla) changes this macroscopic magnetization and proton spins are perturbed (i.e. tipped away from  $B_0$ ). RF pulses are named after their effect on the net magnetization vector, i.e. an RF pulse tilting the net magnetization vector by  $90^\circ$  from the  $z$  direction ( $B_0$ ) into the  $x/y$  plane is called a  $90^\circ$  pulse and a  $180^\circ$  pulse inverts the magnetization vector (i.e.  $z$  to  $-z$ ; [Figure 1a and b](#)). The return of the tipped net magnetic vector to the steady-state equilibrium along the  $B_0$  axis and the decay of net transverse magnetization, respectively, are two independent processes that can be measured [2]; namely spin–lattice ( $T_1$ ) and spin–spin ( $T_2$ ) relaxation time.  $T_1$  and  $T_2$  relaxation times are characteristic for the tissue composition (i.e. local microstructural magnetochemical environment) and provide the main sources of tissue contrast in morphological MRI.

In order to actually quantify  $T_1$  or  $T_2$  relaxation times (i.e.  $T_1$  and  $T_2$  mapping) different clinical MRI protocols are available with specific advantages and disadvantages; the chosen method is often determined by the available MRI hardware, sequence and acquisition time. However, over the years, a plethora of measurement sequences and acronyms have been published,

and the interested reader should refer to an in-depth textbook written particularly for medical doctors [3].

### $T_1$ relaxation time

The gold standard for  $T_1$  measurement, the inversion recovery (IR) technique, first inverts the magnetization in the  $z$  direction using a  $180^\circ$  pulse, which is followed by a waiting time, TI (inversion time), and a successive  $90^\circ$  pulse to initiate data readout with further  $180^\circ$  pulses. This IR preparation module has to be repeated several times by incrementing TI to acquire three to eight data points using a long TR (repetition time; i.e. five to seven times  $T_1$ ), to ensure full relaxation before each inversion pulse, which leads to long overall IR- $T_1$  measurement times ([Figure 1a and c](#)).

The desire for faster  $T_1$  measurement compatible with individual breath-holds has given rise to several efficient methods, the most common being variable flip-angle (VFA) and modified Look-Locker imaging (MOLLI).

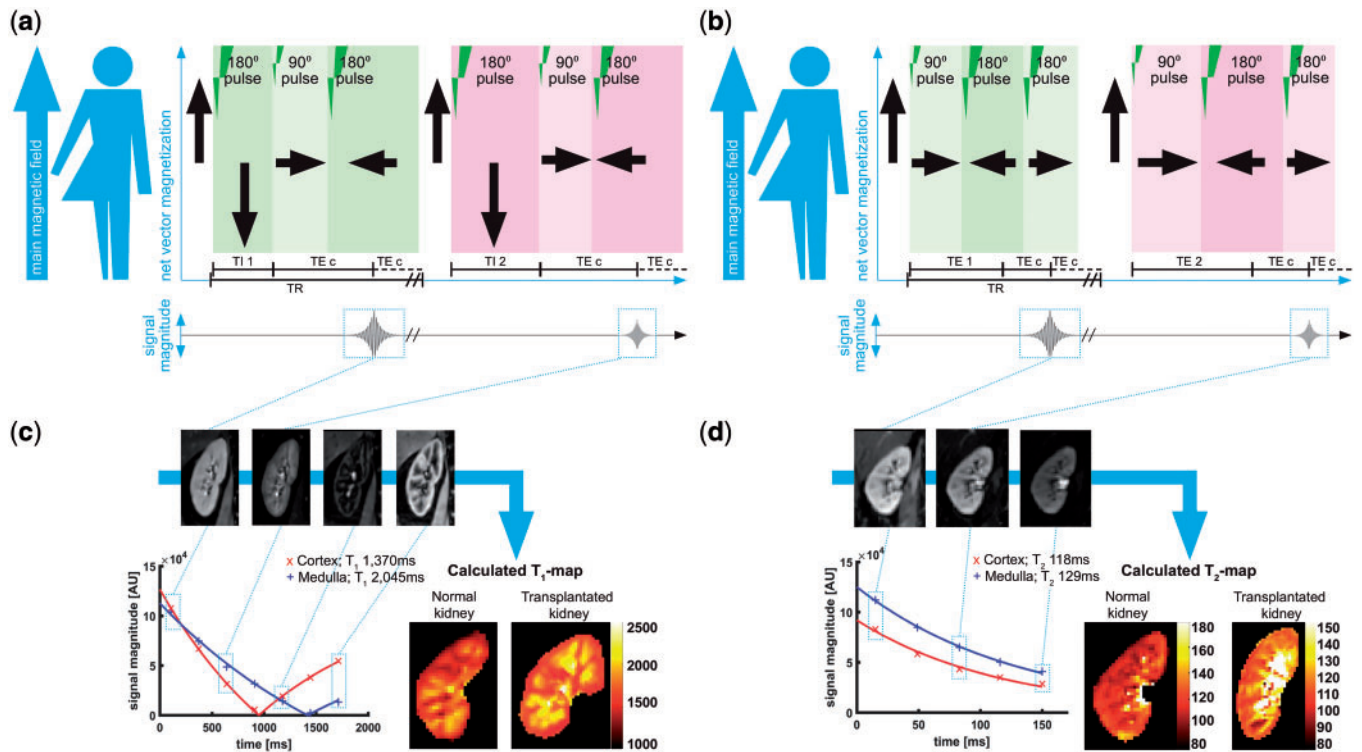
In VFA, two or more spoiled gradient recalled-echo acquisitions with differing excitation pulse flip-angles give rise to signals modulated by  $T_1$  [4]; while substantially faster than IR- $T_1$ , care must be taken before considering VFA to provide quantitative, rather than relative,  $T_1$  measures [5]. VFA measurements are susceptible to  $B_1$  inhomogeneity and thus require additional  $B_1$  mapping. Also, the accuracy of the resulting  $T_1$  depends on the relation of the chosen flip-angles with respect to the observed  $T_1$  range.

The MOLLI sequence and its variants, based on the technique developed in 1970 by Look and Locker [6], sacrifice the requirement of pre-excitation equilibrium to save time and report a modified, shorter, apparent  $T_1$  (often denoted  $T_1^*$ ) derived from repeated efficient sampling of a single excitation pulse. This type of sequence is sufficiently fast, so it is well suited for cardiac imaging, but the comparability between  $T_1$  and  $T_1^*$  is limited [7, 8].

### $T_2$ relaxation time

The most common method to measure  $T_2$  relaxation time is a multi-echo (fast) spin-echo sequence, which first applies a  $90^\circ$  pulse to tilt the magnetization into the  $x/y$  plane and then applies several  $180^\circ$  pulses in the  $x/y$  plane to recover (echo) magnetization and hence enables  $T_2$  estimates from the signal envelope ([Figure 1b and d](#)). This approach is achieved within one TR, which is much faster than  $T_1$  (IR) measurements, and allows full kidney coverage within a few breath-holds.

However,  $T_2$  measurements are sensitive to imperfect slice selection pulse profiles, diffusion, flow and field inhomogeneities [9]. A  $T_2$  preparation module decreases the influence of imperfect slice selection profiles, diffusion and flow. Carr–Purcell–Meiboom–Gill (CPMG) and similar preparations can help to compensate for field inhomogeneities. Therefore  $T_2$  preparations yield more accurate (but slightly higher)  $T_2$  values as compared with a multi-echo spin-echo approach.  $T_2$  preparations are widely used in cardiac imaging to visualize oedema after myocardial infarction [10], and can be performed during free breathing, although image registration prior to  $T_2$  calculation is required. Commonly at least three source images with



**FIGURE 1:** Simplified illustration of the quantification of  $T_1$  (a, c), and  $T_2$  (b, d) relaxation time measurements in the cortex (red) and medulla (blue). The illustration on the left (a, b) shows the patient lying inside the MRI scanner (view from above). The main magnetic field ( $B_0$ ) is in the foot–head direction. The static magnetic field causes some nuclear spins to align parallel with  $B_0$ , which is illustrated with the first big black arrow in the graphic next to it. (a) Simplified sequence diagram for  $T_1$  mapping. The gold standard for  $T_1$  relaxation time measurements is initiated by a  $180^\circ$  pulse (IR). As a consequence, the net magnetization is tilted in the  $z$  direction (from left to right; first grey arrow). Thereafter a waiting time is applied, TI 1 (time of inversion), which ends after the application of a  $90^\circ$  pulse, so that the net magnetization is tilted in the  $x/y$  plane and the readout with constant time of echo (TE c) begins. After a long time of repetition (TR) the next measurement begins; however, the waiting time is longer (TI 2). The graphic below shows the acquired signal, which shows a stronger signal for the first measurement and a weaker signal for the second measurement (see dashed boxes). (b) Simplified sequence diagram for  $T_2$  mapping. The most commonly used protocol is initiated by a  $90^\circ$  pulse and a  $180^\circ$  pulse, which tilts the net magnetization first into the  $x/y$  plane and thereafter into the opposite direction. This process is differently timed (TE 1 and TE 2). After successive  $180^\circ$  pulses the readout begins with TE c. Below, the acquired signal is shown. Notice the exemplified and reduced signal magnitude of the second signal (dashed boxes). (c) Multiple inversion time acquisitions for  $T_1$  mapping. On the bottom left, the graph shows the measured signal magnitude for each inversion time of the IR sequence. Due to the IR the  $T_1$  signal decays first towards null and recovers afterwards, which can also be depicted in the corresponding images of the native kidney on the top left. The  $T_1$  signal decay curve is used to calculate a color-coded  $T_1$  map (examples of normal and transplanted kidney; colour bar in ms). (d) The graph on the bottom left shows the  $T_2$  signal decay during the multiple echo time acquisition for the  $T_2$  mapping data. Corresponding images of the native kidney is shown on the top left. The  $T_2$  signal decay curve is used to calculate a colour-coded  $T_2$  map (examples of normal and transplanted kidney; colour bar in ms). Figure layout, design, and editing: Karin van Rijnbach, A.d.B., N.P.J. and M.W.; image data acquisition and reconstruction: A.d.B.

different echo times are recommended for accurate  $T_2$  estimation using two- or three-parameter exponential fittings [10–13].

## RENAL $T_1$ MAPPING

### Reference values and physiological modulations

In the early 1980s, renal MRI detected relatively increased  $T_1$  values in the medulla compared with the cortex in healthy subjects. This corticomedullary differentiation (CMD) is presumably caused by the higher free water content, i.e. higher mobility of water molecules, in the medullary tubules and collecting ducts [14, 15]. Additionally Hricak *et al.* [14] reported that hydration and the water balance management of the kidneys are important influencing factors, because  $T_1$  CMD decreases during dehydration (relative cortical  $T_1$  increase) and increases

after rehydration, i.e. forced diuresis [14], but the impact in healthy subjects or patients was never reassessed at 1.5 T and 3T. Another inevitable variation is caused by the increase of  $B_0$  from 1.5 and 3T, as  $T_1$  generally increases. Further variation of renal  $T_1$  values was reported due to different MRI acquisition schemes and breathing strategies [16, 17], even though high interexamination repeatability for single acquisition schemes was proven [18–20]. Therefore no widely accepted reference values are published and the given limitations have to be considered when comparing different studies (Table 1).

**$T_1$  modulation by the inhalation of oxygen and carbogen.**  $T_1$  and  $T_2^*$  relaxation times are modulated by oxygen level changes in the blood and/or tissue, although caused by different mechanisms [21].  $T_2^*$ , i.e. blood oxygen level

**Table 1. Quantitative T<sub>1</sub> studies at 1.5 and 3T**

Author	Year	Subject	Sample size	Group	<i>In vivo</i> repeatability	GFR	Hydration	Respiratory compensation	Sequence	Cortex	Medulla	Other modalities	
<b>1.5T</b>													
Biliml <i>et al.</i> [16]	1993	Healthy	9	—	No	Not measured	None	BH	IR TurboFLASH	966 ± 41	1320 ± 76	—	
Jones <i>et al.</i> [21]	2002	Healthy	9	Normoxia Pure O <sub>2</sub>	No	Not measured	None	BH	IR segmented half Fourier TSE	882 ± 59* 829 ± 70*	1163 ± 118 1159 ± 117	—	
de Bazelaire <i>et al.</i> [11]	2004	Healthy	4	—	No	Not measured	None	BH	IR SS FSE, half Fourier	996 ± 58	1412 ± 58	T <sub>2</sub>	
Lee <i>et al.</i> [22]	2007	Underlying renal disease unknown	10	Mixed: 1 patient with CKD and hypertension; 9 patients with only hypertension; 3 patients had RAS	No	mGFR: SKGFR <sup>99mTc-DTPA</sup>	No fasting. Subjects drank ~300mL prior the MR acquisition and voided	BH	IR trueFISP	1083 ± 149	1229 ± 118	—	
O'Connor <i>et al.</i> [23]	2007	Healthy	5	Normoxia (21% O <sub>2</sub> ) Pure O <sub>2</sub> Carbongen (95% O <sub>2</sub> & 5% CO <sub>2</sub> )	No	Not measured	None	FB	VFA 3D T1w FFE	945 ± 15 <sup>▲▼</sup> 883 ± 9 <sup>▲</sup> 873 ± 22 <sup>▼</sup>	Not measured	—	
O'Connor <i>et al.</i> [24]	2009	Healthy	6	Normoxia (21% O <sub>2</sub> ) Pure O <sub>2</sub> Carbongen (95% O <sub>2</sub> & 5% CO <sub>2</sub> )	No	Not measured	4-h fasting	FB	VFA 3D RF-spoiled T1w FFE	961 ± 48 <sup>l/s</sup> 897 ± 27 <sup>l</sup> 909 ± 35 <sup>s</sup>	Not measured	T <sub>2</sub> *	
Huang <i>et al.</i> [19]	2011	Underlying renal disease unknown Renal allograft	5 7 11 4	Native kidney; eGFR <60 (32 ± 13) Native kidney; eGFR >60 (80 ± 7) eGFR <60 (42 ± 15) eGFR >60 (73 ± 5)	No Yes	MDRD eGFR	4-h fasting	BH	IR SS FSE	1145 ± 216 <sup>a</sup> 995 ± 216 <sup>a</sup> 1231 ± 191 <sup>a</sup> 1051 ± 179 <sup>a</sup>	1392 ± 110 <sup>a</sup> 1057 ± 94 <sup>a</sup> 1387 ± 119 <sup>a</sup> 1621 ± 190 <sup>a</sup> 1439 ± 113 <sup>a</sup>	1389 ± 48 <sup>o</sup> — 1573 ± 132 <sup>o</sup>	—
Bredthardt <i>et al.</i> [25]	2015	Preserved renal function Impaired renal function	10 10	Mean eGFR 101 ± 17 Age-matched volunteers; mean eGFR 75 ± 16 cHF: mean eGFR 73 ± 8 cHF: mean eGFR 38 ± 11	No No	MDRD eGFR	None	TRIG	IR bFFE	1080 ± 68 1030 ± 55	1054 ± 65 <sup>■</sup> Not measured	Not measured ASL	
Chen <i>et al.</i> [17]	2016	Healthy	9	—	No	Not measured	None	BH	MOLLI	1067 ± 79 <sup>□</sup>	1121 ± 102 <sup>■</sup> Not measured	ASL	
Cox <i>et al.</i> [18]	2017	Healthy	8 58 38	—	No	eGFR	2-h fasting	BH	IR SE EPI bFFE	827 ± 50 1024 ± 71 1053 ± 72	1381 ± 95 1272 ± 140	— — 1318 ± 98	

Author	Year	Study	n	Native kidneys	No	Cockcroft-Gault eGFR	None	BH	MOLLI	987 ± 102 <sup>▲</sup>	1428 ± 98 <sup>#</sup>	-	
Peperhove <i>et al.</i> [26]	2018	Healthy	14	Native kidneys	No	Cockcroft-Gault eGFR	None	BH	MOLLI	987 ± 102 <sup>▲</sup>	1428 ± 98 <sup>#</sup>	-	
		LuTx	52	Native kidneys							1058 ± 96 <sup>▼</sup>	1414 ± 101 <sup>▶</sup>	
		Renal allograft	49	Renal allograft							1299 ± 101 <sup>▼▲</sup>	1516 ± 76 <sup>#</sup>	
		Healthy, LuTx and renal allograft mixed	47	eGFR ≥ 90							1058 ± 108	1427 ± 89 <sup>†</sup>	
		Healthy, LuTx and renal allograft mixed	26	eGFR 60-89	No	Cockcroft-Gault eGFR	None	BH	MOLLI	1077 ± 132	1421 ± 123 <sup>°</sup>		
		Healthy, LuTx and renal allograft mixed	16	eGFR 30-59							1273 ± 97	1541 ± 51 <sup>†,°</sup>	
		Healthy, LuTx and renal allograft mixed	18	eGFR 15-29							1297 ± 113	1497 ± 97	
		Healthy, LuTx and renal allograft mixed	8	eGFR < 15							1377 ± 109	1515 ± 45	
		3T											
		de Bazelaire <i>et al.</i> [11]	2004	Healthy	6	-	No	Not measured	None	BH	IR SS FSE, half-Fourier	1142 ± 154	1545 ± 142
Ding <i>et al.</i> [27]	2013	Healthy	7	Normoxia; conventional acquisition	No	Not measured	None	BH	IR HASTE breath	1187 ± 112	1523 ± 116	T <sub>2</sub> *	
Gillis <i>et al.</i> [20]	2014	Healthy	12	MRI 1; eGFR 98 ± 15	Yes	CKD eGFR	6-h fasting	BH	MOLLI	1376 ± 104	1651 ± 86	-	
Li <i>et al.</i> [12]	2015	Healthy	5	-	No	No	None	BH	IR SS FSE	1406 ± 96	1639 ± 80	T <sub>2</sub>	
Chen <i>et al.</i> [17]	2016	Healthy	26	-	No	No	None	BH	MOLLI	1261 ± 86	1676 ± 94	-	
Gillis <i>et al.</i> [28]	2016	Healthy	24	Mean eGFR 100 ± 14	No	CKD-EPI eGFR	None	BH	MOLLI	1194 ± 88	1610 ± 55	-	
Friedli <i>et al.</i> [29]	2016	Renal allograft	29	Renal allograft	No	CKD-EPI eGFR	None	BH	MOLLI	1366 ± 122 <sup>▲</sup>	Not measured	-	
Cox <i>et al.</i> [18]	2017	Healthy	21	Renal allograft	No	CKD-EPI eGFR	None	BH	MOLLI	1550 ± 81 <sup>▲</sup>	Not measured	-	
		Healthy	20	Renal allograft	No	CKD-EPI eGFR	None	BH	MOLLI	1334 ± 57	1473 ± 48	-	
		Healthy	26	Age < 40 a	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	1367 ± 79	-		
		Healthy	25	Age < 40 a	Yes	eGFR	2-h fasting	TRIG	IR BFFE	1124 ± 114	-	T <sub>2</sub> *, DWI, PC, ASL	
		Healthy	13	Age > 40 a	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	1347 ± 65	-		
		Healthy	12	Age > 40 a	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	-	1635 ± 66		
		Healthy	8	Age > 40 a	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	1399 ± 93 <sup>#</sup>	1685 ± 84		
		Healthy	11	Mean eGFR 51 ± 15	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	1530 ± 99 <sup>#</sup>	1726 ± 78	T <sub>2</sub> *, DWI, PC, ASL	
		Healthy	21	Mean eGFR 98 ± 15	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	1367 ± 79	-		
		Healthy	20	Mean eGFR 98 ± 15	Yes	eGFR	2-h fasting	TRIG	IR SE EPI	-	1655 ± 76		

The given T<sub>1</sub> relaxation times of the cortex and medulla are mean ± SD in ms. Patient studies are highlighted in grey. 3D, three-dimensional; <sup>99m</sup>Tc-DTPA, <sup>99m</sup>Tc-diethylene triamine pentaacetic acid; a, year; BFFE, balanced fast field echo; BOLD, blood oxygen level dependant; BH, breath hold; CKD, chronic kidney disease; CO<sub>2</sub>, carbon dioxide; DWI, diffusion-weighted imaging; EPI, echo-planar imaging; e/mGFR, estimated or measured glomerular filtration rate (in mL/min/1.73 m<sup>2</sup>); GE, gradient echo; FFE, fast field echo; FSE, fast spin echo; FLASH, fast low angle shot; FB, free breathing; HASTE, half Fourier acquisition single shot turbo spin echo; LuTx, lung transplantation; MDRD, modification of diet in renal disease; ME, multi-echo; MS, multishot; PC, phase contrast; SE, spin-echo; SS, single shot; T, Tesla; T<sub>1</sub>, spin-lattice relaxation time; T<sub>2</sub>, spin-spin relaxation time; T<sub>2</sub>\*, apparent transverse relaxation time; T<sub>1</sub>w, T<sub>1</sub> weighted; TRIG, triggered MRI acquisition with regards to breathing motion; trueFISP, true fast imaging with steady-state precession; TSE, turbo spin echo. <sup>†</sup>Recalculated and corrected values.

Other symbols refer to the statistical significance within the associated study: <sup>‡</sup>p < 0.0001; <sup>†</sup>p < 0.005; <sup>§</sup>p < 0.008; <sup>#</sup>p < 0.01; <sup>°</sup>p < 0.01; <sup>◀</sup>p = 0.01; <sup>▶</sup>p = 0.01; <sup>■</sup>p = 0.03; <sup>□</sup>p = 0.047; \*p < 0.05.

dependent (BOLD) MRI, associated changes are reviewed by Pruijm *et al.* [30].

To our knowledge, modulations of renal  $T_1$  values during the inhalation of pure oxygen ( $O_2$ ) and carbogen (5% carbon dioxide mixed with 95%  $O_2$ ) were only observed in healthy volunteers. In 2002, Jones *et al.* [21] reported a significant decrease in cortical  $T_1$  values during  $O_2$  inhalation at 1.5T. These findings were confirmed in 2007 and 2009 with an even more pronounced reduction in cortical  $T_1$  values following the inhalation of  $O_2$  and carbogen [23, 24]. In these studies, the lack of a renal hydration protocol [except in O'Connor *et al.* [24]], the free breathing acquisition, the VFA method and dyspnoea during the carbogen inhalation (leading to increased breathing motion), as well as the temporal and spatial acquisition constraints, can be considered as important limitations [23, 24].

The first 3T study was carried out by Ding *et al.* [27] when healthy subjects were evaluated during exposure to normoxia and  $O_2$ . Thereafter a multiparametric renal MRI study evaluated five healthy volunteers who underwent a hyperoxia challenge (~80%  $O_2$ ); again cortical  $T_1$  values decreased, but unlike previous publications, no statistical significance was observed [18].

These studies show that cortical  $T_1$  is sensitive to oxygenation level changes. However, the contribution of vasoconstriction and vasodilatation as well as perfusion changes during  $O_2$  and carbogen inhalation, as well as the evaluation of renal oxygen delivery (ischaemia), were never directly assessed, which could have caused the reported ambiguous findings [18, 27]. Final conclusions regarding medullary  $T_1$  modulations are currently not possible. Last but not least, it has not been clarified yet as to what extent alterations in  $T_1$  reflect tissue and/or blood oxygenation. These questions remain a target for future evaluations.

## Clinical studies

**Renal transplants—early qualitative and semi-quantitative MRI studies.** Imaging of renal transplants in the iliac fossae is less confounded by breathing motion, which enabled renal MRI evaluations in the 1980s [31]. Early qualitative and/or semi-quantitative renal MRI studies revealed a reduced  $T_1$  CMD in acute allograft rejection (AAR), and even diminished  $T_1$  CMD in chronic allograft rejection (CAR) [15, 31–34]. However, acute tubular necrosis (ATN) could not be sufficiently differentiated from AAR [32, 34–36], and even diminished  $T_1$  CMD was reversible in some cases of ATN and AAR [36]. Thus scrutiny of the reduced  $T_1$  CMD linked both oedema and fibrosis to prolonged  $T_1$  values, which partially explains the low specificity of these renal transplant evaluations [37].

Another interesting finding on renal transplant observation was the clearly preserved  $T_1$  CMD during an acute decline in renal function under cyclosporine therapy, which was linked to cyclosporine nephrotoxicity (CN) [32, 34]. However, three successive studies presented ambiguous outcomes [33, 37, 38]. Thereafter, no further research efforts were made, so no final conclusion can be made.

All these envisioned early MRI studies on renal transplants applied field strengths <1.5T, which today are not frequently in

clinical use. However, in contrast to recent MRI evaluations, all of these studies applied histological validation. A low specificity was observed due to different acquisition settings (e.g. vendors and protocols), low reproducibility of the two-point method to calculate  $T_1$  [31] and lack of a standardized patient preparation (e.g. hydration protocol) [14, 15]. In addition, loss of  $T_1$  CMD was reversible after clinical improvement in some cases of ATN and ARR, which could have decreased the specificity further [36]. Therefore recommendations could not advocate qualitative and/or semi-quantitative MRI evaluations over ultrasound and scintigraphy [34].

**Renal transplants—quantitative MRI studies.**  $T_1$  measurements on renal transplants at 1.5T were presented by Huang *et al.* [19] in 2011, when renal transplants and native kidneys with unknown underlying renal disease confirmed the trend of higher cortical and medullary  $T_1$  values in renal transplants. They also achieved a high short-term *in vivo* repeatability ( $\sim\pm 10\%$ ). In addition, strong correlations were observed between estimated glomerular filtration rate (eGFR) and cortical  $T_1$  in both groups (native cortex:  $r = -0.83$ ,  $P = 0.0001$ ; transplant cortex:  $r = -0.80$ ,  $P = 0.0017$ ), but medullary  $T_1$  values only significantly correlated with eGFR in the transplant group ( $r = -0.94$ ,  $P < 0.0001$ ) [19].

The second quantitative  $T_1$  assessment of renal transplant was presented by Friedli *et al.* [29]. A total of 29 patients underwent a multiparametric MRI approach at 3T, including a validation against histological samples. With regard to  $T_1$ , only  $T_1$  CMD showed a moderate correlation with renal interstitial fibrosis ( $R^2 = 0.29$ ,  $P < 0.001$ ) and eGFR ( $R^2 = 0.22$ ,  $P < 0.05$ ). No correlation was established between  $T_1$  values and cellular inflammation [29]).

In 2018, renal  $T_1$  was evaluated in 49 renal transplant patients, 52 patients after lung transplantation (LuTx; native kidneys) and 14 healthy volunteers [26]. Their aim was to assess acute kidney injury (AKI) after LuTx (reported incidence ~60%), and after a 3- and 6-month follow-up.  $T_1$  CMD was significantly decreased and mean cortical and medullary  $T_1$  were significantly higher in renal transplants compared with healthy volunteers and the LuTx group ( $P < 0.001$ ). However,  $T_1$  CMD was also reduced in the LuTx group compared with volunteers ( $P < 0.05$ ), which was linked to the incidence of AKI after LuTx. All patients and healthy volunteers were further grouped according to Kidney Disease Outcomes Quality Initiative (KDOQI) stages. Remarkable were the significantly lower cortical  $T_1$  values in subjects with  $eGFR \geq 60$  mL/min/1.73 m<sup>2</sup> as compared with  $<60$  mL/min/1.73 m<sup>2</sup> and that cortical  $T_1$  negatively correlated ( $r = -0.642$ ,  $P < 0.001$ ) and  $T_1$  CMD positively correlated ( $r = 0.542$ ,  $P < 0.001$ ) with eGFR for all participants. In contrast, medullary  $T_1$  showed only a weak correlation with eGFR ( $r = -0.341$ ,  $P < 0.001$ ). During the 3- and 6-month follow-up, cortical  $T_1$  and  $T_1$  CMD exhibited a significant correlation with eGFR ( $P < 0.001$  and  $< 0.01$ , respectively) in the LuTx and renal transplantation groups [26].

In summary, we identified only three quantitative  $T_1$  studies on renal allografts at 1.5 and 3T. In contrast to early qualitative and semi-quantitative MRI studies, only one quantitative study applied a histological validation, in which it was shown that

**Table 2. Quantitative T<sub>2</sub> studies at 1.5 and 3T**

Author	Year	Subject	Sample size	Group	In vivo repeatability	GFR	Hydration	Respiratory compensation	Sequence	Cortex	Medulla	Other modalities
<b>1.5T</b>												
de Bazelaire <i>et al.</i> [11]	2004	Healthy	4	—	No	Not measured	None	BH	SE T <sub>2</sub> prep	87 ± 4	85 ± 11	T <sub>1</sub>
Zhang <i>et al.</i> [45]	2011	Healthy	4	Day 1 Day 2	Yes	Not measured	None	BH	2D ME TSE	112 <sup>†</sup> 112 <sup>†</sup>	137 <sup>†</sup> 143 <sup>†</sup>	T <sub>2</sub> *
Mathys <i>et al.</i> [46]	2011	Healthy	6	—	No	TUC	2-h fasting	FB	ME SE	125 ± 7 <sup>#,o</sup>	—	T <sub>2</sub> *
		Renal allograft	6	GFR >40 GFR <40	No	TUC	2-h fasting	FB	ME SE	147 ± 13 <sup>#</sup> 150 ± 20 <sup>o</sup>	—	T <sub>2</sub> *
<b>3T</b>												
de Bazelaire <i>et al.</i> [11]	2004	Healthy	6	—	No	Not measured	None	BH	SE T <sub>2</sub> prep	76 ± 7	81 ± 8	T <sub>1</sub>
Li <i>et al.</i> [12]	2015	Healthy	5	—	No	Not measured	None	BH	CPMG T <sub>2</sub> prep	121 ± 5	138 ± 7	T <sub>1</sub>
Franke <i>et al.</i> [47]	2017	Healthy	3	—	No	Not measured	None	—	ME GE SE	132 ± 6 <sup>#,◀</sup>	—	—
		ADPKD	3	TKV <300 mL TKV 300–400 mL TKV >400 mL	No	Not measured	None	—	ME GE SE	417 ± 65 <sup>*,*</sup> 592 ± 231 <sup>◀</sup> 669 ± 170 <sup>*,▼</sup>	—	—

The given T<sub>2</sub> relaxation times of the cortex and medulla are mean ± SD in ms. Patient studies are highlighted in grey.

2D, two dimensional; BH, breath-hold; FB, free breathing; GE, gradient echo; ME, multi-echo; prep, preparation; T, Tesla; T<sub>1</sub>, spin-lattice relaxation time; T<sub>2</sub>, spin-spin relaxation time; T<sub>2</sub><sup>\*</sup>, apparent transverse relaxation time; T<sub>1w</sub>, T<sub>1</sub> weighted; TKV, total kidney volume; TUC, timed urine collection; TSE, turbo spin echo.

<sup>†</sup>Recalculated: reported values in mean ± SD; R<sub>2</sub> day 1: 8.9 ± 0.6s<sup>-1</sup> (cortex) and 7.3 ± 0.7s<sup>-1</sup> (medulla); day 2: 8.9 ± 0.6s<sup>-1</sup> (cortex) and 7.0 ± 0.7s<sup>-1</sup> (medulla).

Other symbols refer to the statistical significance within the associated study:

◀\*P < 0.001; #P < 0.01; °P < 0.05.

state-of-the-art  $T_1$  measurements, i.e.  $T_1$  CMD, could be used to assess renal interstitial fibrosis in allografts [29]. Another important finding was that  $T_1$  values were sensitive to presumable AKI alterations in the context of post-LuTx [26]. However, the specificity of renal MRR regarding AAR, CAR, ATN or drug-induced toxicity was not further assessed or improved. Furthermore, these studies show that  $T_1$  mapping has the potential to estimate renal function.

**Non-invasive assessment of renal function.** The first quantitative  $T_1$  measurements on patients at 1.5T were published in 2007 [22]. A small and unbalanced group was primarily enrolled for the evaluation of a renal artery stenosis: one patient with CKD and hypertension and nine patients with hypertension alone. A loose hydration protocol was applied before the MRI acquisition, and afterwards all patients underwent a  $^{99m}\text{Tc}$ -diethylene triamine pentaacetic acid renography to measure the single-kidney GFR (SKGFR). A significant correlation was depicted only between cortical  $T_1$  values and the SKGFR ( $r = -0.5$ ,  $P = 0.03$ ) [22].

In 2015 the association between cortical  $T_1$ , renal perfusion (from arterial spin labeling (ASL); see also Odudu *et al.* [39]) and eGFR in patients with chronic heart failure (HF) and control subjects with different levels of renal impairment was evaluated [25]. Renal perfusion was similar in chronic HF patients with and without renal impairment, but cortical  $T_1$  showed a significant correlation with eGFR ( $r = -0.41$ ,  $P = 0.015$ ), which reflects the potential to assess CKD. Chronic HF patients had significantly higher cortical  $T_1$  compared with all control subjects, and chronic HF patients with renal impairment had significantly higher cortical  $T_1$  compared with chronic HF patients without renal impairment [25].

After the ASL reproducibility study of Gillis *et al.* in 2014 [20], a follow-up study evaluated renal perfusion and cortical  $T_1$  in healthy volunteers and CKD patients with different etiologies at 3T. Significantly higher cortical  $T_1$  values were found in CKD patients and a correlation between cortical  $T_1$  and eGFR was observed ( $r = -0.58$ ,  $P < 0.001$ ) [28].

One year later a multiparametric renal MRI study assessed  $T_1$  in healthy subjects and CKD patients with various renal diseases after a short fasting period ( $>2\text{h}$ ) at 3T [18]. Compared with volunteers, CKD showed significantly higher cortical  $T_1$ , and  $T_1$  CMD was reduced ( $P < 0.01$ ). They achieved an interscan coefficient of variation of  $<2.9\%$  and high intraclass correlation for the cortex and medulla (0.848 and 0.997, respectively, using spin-echo echo-planar imaging) [18].

As previously envisioned also, three renal transplant studies assessed the correlation of  $T_1$  values and the renal function at 1.5 and 3T (see above) [19, 26, 29].

In summary, the envisioned studies show that the degree of renal impairment correlates moderately to strongly with cortical  $T_1$  and  $T_1$  CMD in CKD with various renal diseases [18, 22, 28], renal transplants [19, 26, 29], and chronic HF patients [25]. These findings are also in line with some qualitative assessments in the 1990s [40, 41], but not with all [42], due to the fact that renal  $T_1$  values are modulated by many confounders, such as the degree of fibrosis [29], comorbidities (e.g. liver cirrhosis) [43, 44], the acquisition protocol (e.g. breathing motion) and

fastening and hydration level [14], which all together seem to be responsible for the accomplished correlations in the envisioned quantitative studies at 1.5 and 3T. To our knowledge, only one study correlated renal  $T_1$  values with measured GFR [22]. It should be noted that adequate patient preparation (e.g. hydration protocol, medication intake), patient selection in the context of comorbidities and acquisition protocols (e.g. triggered breath-hold) together with reference measurement of the renal function can improve  $T_1$  renal function correlations, which advocates for further research in this field.

## RENAL $T_2$ MAPPING

### Reference values and physiological modulations

In healthy subjects, medullary  $T_2$  is consistently longer than cortical  $T_2$ . As previously envisioned, Hricak *et al.* [14] evaluated the effect of fasting and hydration and showed that  $T_2$  CMD decreased during hydration (i.e. forced diuresis), but these findings were never re-evaluated. Additional variation can also be found due to the increase in  $B_0$  from 1.5 and 3T, which is accompanied by a general decrease in  $T_2$ , and by the fact that different MRI acquisitions and breathing strategies report unequal values. But for healthy subjects a high day-to-day repeatability was shown by a multi-echo spin-echo method with a mean variability of  $<4\%$  for both cortex and medulla at 1.5T [45].

Closely linked to  $T_2$  is  $T_2'$ , which is thought to reflect tissue oxygenation [45, 46]. For measurement of  $T_2'$ , both  $T_2$  and  $T_2^*$  are required.  $T_2^*$ , i.e. renal BOLD MRI, is discussed by Pruijm *et al.* in this issue [30].

These variations have to be considered when comparing different studies (Table 2).

### Clinical studies

In the 1980s renal transplants were evaluated regarding  $T_2$ , and MRI was shown to be useful to identify fluid collections in necrotic transplant, perinephric lymphoceles and haematoma [31].

To our knowledge, the first quantitative clinical, i.e. renal transplant, study on  $T_2$  values at 1.5Tesla (T) was reported in 2011. One of two  $T_2$  acquisition protocols identified a significant increase in cortical  $T_2$  in 15 renal transplants compared with 6 healthy subjects. However, no significant difference was observed with regards to the allograft function [46].

In 2017, whole kidney  $T_2$  values in animals with juvenile cystic kidneys and nine autosomal dominant polycystic kidney disease (ADPKD) patients were reported. A strong significant increase in  $T_2$  values was seen in early-stage ADPKD patients compared with healthy volunteers. Based solely on  $T_2$  values, early-stage ADPKD patients with a kidney volume  $<300\text{ mL}$  could be distinguished from healthy volunteers, which was not possible based on total kidney volume (TKV) [47].

In summary, human *in vivo* measurements of renal  $T_2$  are relatively scarce. Therefore no final conclusion can be made regarding renal function estimation or renal transplant assessments. Nevertheless, interesting findings were obtained, which clearly advocate for future research. Early-stage ADPKD



patients could benefit from the  $T_2$  evaluations and the potentially improved assessment of early-disease progression compared with TKV [47]. This might be of special interest in the evaluation of novel therapeutic agents such as tolvaptan. The assessment of AKI in the context of ischaemia reperfusion injury, e.g. induced kidney damage during renal allograft surgery, also seems to be a potential application for  $T_2$ , as *in vivo* measurements were shown to be feasible [46]. Animal studies have shown that  $T_2$  is sensitive to ischaemia–reperfusion injury [48, 49]. During initial ischaemia,  $T_2$  decreases, probably due to deoxygenation, followed by an increase during reperfusion [50]. In the longer term, an elevation of  $T_2$  that is more pronounced in the medulla compared with the cortex has been found [51, 52], which was attributed to consecutive inflammation and oedema ( $T_2$  increase) [50–52]. Human studies are necessary to determine whether the  $T_2$  changes following AKI can predict the recovery of renal function.

## DISCUSSION

In recent decades, quantitative renal  $T_1$  and  $T_2$  mapping have been shown not only to be feasible, but also to provide non-invasive valuable information regarding renal structure and function in healthy, AKI, CKD, renal transplant and ADPKD patients at 1.5 and 3T (Tables 1 and 2).

Renal  $T_1$  has been shown to be modulated by hydration and, in particular, cortical  $T_1$  was sensitive to oxygenation.  $T_1$  CMD is a potential candidate biomarker to assess AAR, CAR, ATN, CN, fibrosis and renal function. Renal  $T_2$  was measured in only a few studies but showed the potential to evaluate renal transplants and to improve the diagnosis and progression of early-stage ADPKD.

However, the variation in  $T_1$  and  $T_2$  values is large, mainly due to the great diversity of the MRR methods applied, but also due to physiological (e.g. water balance management during fasting and forced diuresis) and pathological alterations (e.g. fibrosis) of the renal parenchyma. In virtually all renal diseases, renal function and microstructure are altered together, and this review on  $T_1$  and  $T_2$  unveiled the high sensitivity towards each of these processes as well as the complicated interpretation of the acquired data due to the low specificity.

In conclusion, currently available data suggest that the full potential of renal  $T_1$  and  $T_2$  mapping has not yet been tapped and adequate patient selection, with regard to comorbidities, alongside technical and physiological standardization, will significantly increase the specificity of renal MRR. On route towards renal  $T_1$  or  $T_2$  mapping as a biomarker it will be necessary to validate renal MRR against widely accepted reference measurements (e.g. nuclear medicine evaluations) as well as against histological findings, when possible. Last but not least, the integration of different quantitative renal MRI data into a multi-parametric approach will likely enable us to gain the best insight into renal pathophysiology. The COST Action PARENCHIMA ([www.renalMRI.org](http://www.renalMRI.org)) is working on standardization of multiparametric renal MRI techniques to tackle these challenges.

## SUPPLEMENTARY DATA

Supplementary data are available at [ndt online](http://ndt-online).

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## CONFLICT OF INTEREST STATEMENT

None declared.

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