Renal blood oxygenation level-dependent magnetic resonance imaging to measure renal tissue oxygenation: a statement paper and systematic review

Menno Pruijm¹, Iosif A. Mendichovszky², Per Liss³, Patricia Van der Niepen⁴, Stephen C. Textor⁵, Lilach O. Lerman⁵, C.T. Paul Krediet⁶, Anna Caroli⁷, Michel Burnier¹ and Pottumarthi Vara Prasad⁸

¹ Service of Nephrology and Hypertension, Department of Medicine, University Hospital of Lausanne (CHUV), Lausanne, Switzerland, ²Department of Radiology, Cambridge University Hospitals NHS Foundation Trust, Addenbrooke’s Hospital, Cambridge, UK, ³Department of Surgical Sciences, Section of Radiology, Uppsala University, Uppsala, Sweden, ⁴Department of Nephrology and Hypertension, Universitair Ziekenhuis Brussel (VUB), Brussels, Belgium, ⁵Division of Nephrology and Hypertension, Mayo Clinic, Rochester, MN, USA, ⁶Department of Internal Medicine, Division of Nephrology, Academic Medical Center at the University of Amsterdam, Amsterdam, The Netherlands, ⁷Medical Imaging Unit, Bioengineering Department, IRCCS Istituto di Ricerche Farmacologiche Mario Negri, Bergamo, Italy and ⁸Department of Radiology, NorthShore University HealthSystem, Evanston, IL, USA

Correspondence and offprint requests to: Menno Pruijm; E-mail: menno.pruijm@chuv.ch; Twitter handle: @renalMRI

ABSTRACT

Tissue hypoxia plays a key role in the development and progression of many kidney diseases. Blood oxygenation level-dependent magnetic resonance imaging (BOLD-MRI) is the most promising imaging technique to monitor renal tissue oxygenation in humans. BOLD-MRI measures renal tissue deoxyhaemoglobin levels voxel by voxel. Increases in its outcome measure R2* (transverse relaxation rate expressed as per second) correspond to higher deoxyhaemoglobin concentrations and suggest lower oxygenation, whereas decreases in R2* indicate higher oxygenation. BOLD-MRI has been validated against micropuncture techniques in animals. Its reproducibility has been demonstrated in humans, provided that physiological and technical conditions are standardized. BOLD-MRI has shown that patients suffering from chronic kidney disease (CKD) or kidneys with severe renal artery stenosis have lower tissue oxygenation than controls. Additionally, CKD patients with the lowest cortical oxygenation have the worst renal outcome. Finally, BOLD-MRI has been used to assess the influence of drugs on renal tissue oxygenation, and may offer the possibility to identify drugs with nephroprotective or nephrotoxic effects at an early stage. Unfortunately, different methods are used to prepare patients, acquire MRI data and analyse the BOLD images. International efforts such as the European Cooperation in Science and Technology (COST) action ‘Magnetic Resonance Imaging Biomarkers for Chronic Kidney Disease’ (PARENCHIMA) are aiming to harmonize this process, to facilitate the introduction of this technique in clinical practice in the near future. This article represents an extensive overview of the studies performed in this field, summarizes the strengths and weaknesses of the technique, provides recommendations about patient preparation, image acquisition and analysis, and suggests clinical applications and future developments.

Keywords: BOLD-MRI, chronic kidney disease, functional MRI, kidney, renal artery stenosis

INTRODUCTION

Renal tissue ischemia is one of the main mechanisms leading to acute kidney injury and mounting evidence suggests that renal tissue hypoxia is a unifying factor in the development and progression of chronic kidney disease (CKD), irrespective of its cause [1]. Assessing renal tissue oxygenation in humans is therefore highly relevant to clinical practice. Blood oxygenation level-dependent magnetic resonance imaging (BOLD-MRI) has become a widely used method to estimate renal tissue oxygenation, although its use is thus far largely restricted to the research setting.

This review paper, written by the European Cooperation in Science and Technology (COST) action ‘Magnetic Resonance Imaging Biomarkers for Chronic Kidney Disease’ (PARENCHIMA) group, provides an overview of the technique and pitfalls of BOLD-MRI, as well as a summary of all renal BOLD-MRI studies performed in humans so far. These studies...
are summarized in tables under ‘Supplementary data’. This review paper also identifies gaps in knowledge and provides recommendations about patient preparation, image acquisition and analysis, and possible clinical applications.

**METHODODOLOGY**

A comprehensive search of PubMed and Ovid MEDLINE was performed using the following pre-defined MeSH terms: blood, oxygenation, MRI and kidney (renal). Review articles or studies performed in animals were not included, except those that validated BOLD-MRI against directly measured renal tissue oxygenation (micropuncture). Renal cancer studies were excluded. Only original articles written in English and published in peer-reviewed journals before the 31 December 2017 were selected and divided into five categories: validation of the technique, CKD, renal artery stenosis, transplantation and dietary/drug intervention. Each study is summarized in Excel tables (Supplementary data).

**PHYSIOLOGY OF RENAL TISSUE OXYGENATION AND PRINCIPLES OF BOLD-MRI**

Renal tissue oxygenation is the net balance between oxygen delivery and demand. Delivery depends on renal blood flow (RBF), haemoglobin levels, blood pH and oxygen saturation, whereas oxygen demand depends on tissue metabolic requirements and the oxygen-consuming active tubular transport of electrolytes [2]. Decreases in global RBF do not necessarily lead to a reduction in oxygen delivery due to the presence of substantial arterio-venous shunting of oxygen that can vary according to metabolic needs. Inversely, an increase in RBF does not necessarily increase tissue oxygenation since it also leads, at constant filtration fraction (FF), to a higher glomerular filtration rate (GFR), higher amounts of filtered sodium and thus increased metabolic demands [3]. Together, these mechanisms tightly control renal tissue oxygenation. Despite the fact that each kidney receives ~10% of the cardiac output, local pO2 levels are low. Micropuncture studies in animals have measured tissue pO2 of around 40–50 mmHg in the cortex and as low as 10–20 mmHg in the inner medulla [4]. These pO2 levels are at the steep part of the oxygen dissociation curve for haemoglobin, meaning that small changes in local pO2 lead to relatively large changes in the percentage of saturated haemoglobin.

BOLD-MRI uses the specific paramagnetic properties of deoxyhaemoglobin to assess renal tissue oxygenation. Its outcome measure is the so-called R2* value, which varies inversely with local oxygenation. This apparent relaxation rate R2* (or decay rate, defined as 1/T2* and expressed as per second) is measured voxel by voxel in each kidney by BOLD-MRI. The R2* value of each voxel is influenced by microscopic inhomogeneities in the static magnetic field, largely due to the effect of deoxyhaemoglobin. As such, the decay rate R2* in a voxel will be enhanced when the local deoxyhaemoglobin concentration is increased. Hence, assuming that blood pO2 is in equilibrium with tissue pO2, low R2* values indicate high tissue oxygenation, while high R2* values indicate low tissue oxygenation [5].

BOLD-MRI is performed without the administration of contrast agents, which makes it attractive for patients with decreased renal function.

**VALIDATION OF KIDNEY BOLD-MRI**

The aim of BOLD-MRI is to measure renal tissue oxygenation. The actual ‘gold standard’ for tissue oxygenation is direct measurement with micro-electrodes placed in the renal cortex and medulla. For obvious reasons, this is not possible in humans, and comparison with the gold standard has only been performed in animals.

In 1996, Prasad et al. published their landmark paper on the feasibility of the evaluation of intrarenal oxygenation by BOLD-MRI [6]. They studied seven healthy young subjects and found, as expected, a decreased R2* (indicating increased pO2) during administration of furosemide; this observation was in line with prior research using micro-electrodes that had shown an increase in medullary pO2 due to the furosemide-induced reduction in oxygen-consuming active electrolyte transport. In the subsequent years numerous authors confirmed the effects of furosemide in pigs [7], healthy humans [8] and patients [9].

Only a decade later (in 2005) Pedersen et al. took up the challenge to quantitatively validate the technique [7]. They compared BOLD-MRI measurements in renal cortex and medulla to those of pO2 electrodes inserted in the contralateral kidney of eight anaesthetized pigs. They were able to establish a linear relationship between measured pO2 and R2* over the pO2 range of 0.2–10 kPa in both the cortex and the medulla. For the cortex this resulted in a pO2 [kPa] = R2* [1/s] – 21.2 and for the medulla pO2 = R2* – 25/1.7. A study performed by Warner et al. (2011) tested the acute effects of furosemide and acetazolamide in four series of pigs (total n = 21) with consecutive measurements of cortical and medullary oxygenation by BOLD-MRI and Clark electrodes 5–7 days apart [10]. They also found a linear relationship, namely pO2 [kPa] = R2* [1/s] – 20/1.3 for the cortex and for the medulla pO2 = R2* – 20/1.5. The reported pO2 range was narrower than that of Pedersen et al., i.e. 3.1–7.6 kPa, which may explain some of the differences in calculated coefficients and Y-intercepts. Up to now, the two independent studies by Pedersen et al. and Warner et al. serve as the most cited calibration data on renal BOLD-MRI [7, 10]. Recently, Zhang et al. provided a method to relate R2* measurements to blood and tissue pO2 [11].

So far, three studies have specifically assessed the reproducibility of BOLD-MRI in humans. All used the classical region-of-interest (ROI) technique for image analysis (see below). Li et al. showed a ~12% coefficient of variation (CV) for day-to-day measurements in eight healthy subjects at 1.5 T [8]. Simon-Zoula et al. evaluated the reproducibility in 18 healthy volunteers during three subsequent MRI acquisition on a single day, with short interruptions between sessions [12]. They reported CVs of R2* values of 3% and 4% in the medulla and cortex, respectively. Khatir et al. reported CVs of 8% for cortex and medulla in 11 CKD patients examined at 1–2 week intervals, as compared with 5.6% and 3.6% in controls [13].
**PATIENT PREPARATION**

Many factors may alter cortical and medullary \( R^2^* \) values and should be taken into account. Prasad et al. were the first to show that hydration status influences the BOLD signal. They reported decreases up to 30% in medullary \( R^2^* \) values after an acute oral water load of 20 mL/kg in young healthy volunteers [14]. Most BOLD-MRI studies therefore standardize water intake, although approaches differ: some perform BOLD-MRI after an overnight or 4-h fast [15] whereas others prescribe a constant hourly water intake, in general of 1.5 mL/kg/h [16].

A theoretical disadvantage of hydration is that differences in \( R^2^* \) may be ‘washed out’ by hydration, but this only occurs at extreme water intake (\( \sim 1 \) L intake in 15 min) [16]. The clear advantage of hydration is its feasibility in patients; the authors therefore prefer standardized hydration protocols.

Dietary salt intake also influences \( R^2^* \) values. This has been demonstrated in a cross-over study that included young normotensive and untreated hypertensive male volunteers [17]. After 5 days of a low-salt diet, medullary \( R^2^* \) values were significantly lower than after a 5 day high-salt diet. Despite the availability of these data in the literature, less than half of the published studies have controlled salt intake.

Furthermore, as expected, renal tissue oxygenation is influenced by carbogen and oxygen breathing [18].

Some questions concerning patient preparation remain unanswered. For example, no study has investigated whether there is a circadian rhythm in the BOLD signal. Besides, plasma osmolality, pH and temperature (affecting \( O_2 \) dissociation), iron status (affecting tissue para-magnetic properties) and carbon monoxide levels (tobacco use) could all potentially affect the BOLD-MRI signal. None of the latter factors has been studied for this purpose yet. However, subjects are advised to avoid smoking prior to the MRI examination and a delay of 1–2 weeks is usually maintained in case of fever or intravenous (IV) iron perfusion.

**MRI ACQUISITION**

The most common techniques used for human renal BOLD-MRI experiments are single shot echo planar imaging (EPI) [6] and multiple gradient echo (GRE) imaging [19]. Almost all measurements are expiratory breath-hold acquisitions because \( T^2^* \)-weighted renal MRI is prone to motion artefacts. To date, all published studies of the human kidney have been performed at either 1.5 or 3 T field strengths. The most recent publications show preference for 3 T due to improved signal-to-noise ratio (SNR) and better \( T^2^* \)-weighting secondary to \( T^2^* \) shortening. \( R^2^* \) increases with increasing field strength. For example, the \( R^2^* \) value of the healthy medulla at 3 T is almost twice the one acquired at 1.5 T. However, artefacts caused by bulk magnetic susceptibility (BMS) effects and water-fat chemical shift are also magnified at \( T^2^* \) field strengths, but remain at an acceptable level when compared with 1.5 T.

Single shot EPI is a robust approach to reduce motion artefacts and it offers the highest temporal resolution (<100 ms). Unfortunately, EPI is sensitive to BMS artefacts caused by the shape of the human body (organs) and by magnetic susceptibility variations within the body, sometimes leading to image blurring and geometrical distortions. Image distortions are amplified at higher magnetic field strengths, particularly in regions with poor magnetic field homogeneity. To address these challenges, two-dimensional multiple gradient recalled echo (2D mGRE) sequences with signal collection at multiple echoes after each excitation pulse have been introduced and are the most frequently used for renal \( R^2^* \) mapping. Typically, 8, 12 or 16 individual \( T^2^* \)-weighted images of a single slice, corresponding to different echoes, are measured within a single breath-hold of \( \sim 17–20 \) s. However, multislice 2D mGRE as well as 3D mGRE were also successfully implemented during a breath-hold of <23 s [20]. A detailed discussion of the acquisition parameters is beyond the scope of this article, and partly based on expert opinion. The acquisition parameters [repetition time (TR), echo time (TE) interval, acquisition voxel size, flip angle and receiver bandwidth] have to be carefully chosen to maintain the minimum SNR >2 necessary to avoid errors in \( T^2^* \) fitting as the result of noise at higher TE. For more details, see Supplementary data, Table S1. It should be emphasized that the SNR >2 is critical for the robust assessment of the medulla. Both 2D mGRE and as EPI sequences suffer from magnitude and phase variations between odd and even echoes due to different phases of water and fat signal components in renal tissue (in- and out-phase effects). However, these artefacts are usually satisfactorily suppressed during mono-exponential least square fitting of \( T^2^* \).

Despite these improvements, it remains difficult to overcome all scanner-related and other non-renal factors that influence the BOLD signal. This hampers the comparison of absolute \( R^2^* \) values across sites. One way to partly overcome this limitation is to perform a dynamic test (IV furosemide for example) and assess the change in \( R^2^* \) signal. Another way is the use of analysis techniques that are less dependent of the absolute \( R^2^* \) values and focus more on relative changes in \( R^2^* \) (12-layer concentric objects—TLCO and fractional tissue hypoxia technique).

**IMAGE ANALYSIS**

Four major methods have been used thus far to analyse the BOLD images and report the \( R^2^* \) values: ROI technique, TLCO or ‘onion peel’ technique, the fractional tissue hypoxia and the compartmental method. The basic principles, strengths and weaknesses of each technique are summarized in Table 1, and graphical examples are shown in Figure 1.

Several studies [9, 19, 21–22] have specifically assessed the inter-observer variability (image analysis of the same subject by two or more different observers) of these techniques. Piskunowicz et al. reported inter-observer CVs with the ROI technique of 3.6% and 6.8% of cortical and medullary \( R^2^* \) values in healthy volunteers, versus 5.7% and 12.5% in CKD. With the semi-automated TLCO technique, the same research group reported much lower CVs of 2.2%, 2.0% and 3.1% in healthy, mild CKD and severe CKD groups [21]. Cox et al. reported CVs of 4.1% in healthy volunteers with the compartmental method [23].

Most authors agree that the ROI technique is reproducible and easy to use in subjects with (near) normal renal function (GFR), but unreliable in advanced CKD, due to its dependence of the human eye to distinguish cortex from medulla. In CKD, the TLCO technique has the lowest inter-observer variability, whereas the fractional hypoxia technique is mostly used in patients suffering from renal artery stenosis. The compartmental method needs confirmation in CKD patients.
Chronic kidney disease

According to the chronic hypoxia hypothesis, renal tissue hypoxia is a unifying pathway in the development and progression of CKD, irrespective of its cause [1]. Many animal studies support this hypothesis and intra-renal microelectrodes have indeed measured decreased local tissue \( \text{pO}_2 \) in several experimental models of CKD [24, 25].

Initially, numerous BOLD-MRI studies in humans failed to confirm the hypoxia hypothesis [16, 26]. This was probably due to the lack of standardized patient preparation, suboptimal...
study ROI technique that is characterized by high inter-observer variability in CKD. Recent studies performed under standardized conditions and high field strength (3.0 T) have reported higher cortical $R2^*$ values in CKD compared with controls [9, 27]. Of note, most studies have not measured RBF, which complicates the interpretation of the results. In this respect, van der Bel et al. have shown that $R2^*$ values correlate strongly with the FF (GFR/effective renal plasma flow) in healthy volunteers, and they recommend performing simultaneous BOLD-MRI and phase contrast MRI to measure RBF in the renal artery [28].

This field was also hampered by the lack of prospective studies, but this has recently changed with the publication of Pruijm et al. In their cohort of 112 CKD patients, 47 hypertensives and 24 healthy controls, CKD patients with high $R2^*$ values (>90th percentile) had a faster yearly estimated GFR decline and more often needed renal replacement therapy than those with lower $R2^*$ values [29]. This study strongly supports the hypoxia hypothesis and opens the door to more widespread implementation of renal BOLD-MRI in the clinical care for CKD patients.

Renal artery stenosis and arterial hypertension

The decrease in RBF that characterizes renal artery stenosis (RAS) makes BOLD-MRI particularly suitable to probe tissue oxygenation in the post-stenotic kidney. Studies in swine models of acute [30] and chronic [10] reductions in RBF have confirmed that BOLD-MRI is capable of detecting changes in renal oxygenation that were independently confirmed using implanted oxygen electrodes. Studies in human subjects with >60% obstruction of the renal artery have enabled the determination of the haemodynamic significance of the stenosis. Interestingly, cortical and medullary $R2^*$ values were preserved in many patients with moderate obstructions of the renal artery, despite a fall in stenotic kidney RBF and GFR by 30–35% [31]. This finding is consistent with observed adaptability of human kidneys to moderate reductions of RBF. Repeat measurements 3 months later showed that their $R2^*$ values remained highly stable [32]. Blunted responses to the loop diuretic furosemide in these kidneys suggested that preservation of tissue oxygenation was associated with attenuated active solute transport activity and thereby oxygen consumption.

However, kidneys distal to severe luminal obstructions showed more pronounced hypoxia [33] with increased tissue injury, interstitial fibrosis and inflammation [34]. The degree of renal hypoxia by BOLD-MRI in those stenotic kidneys correlated directly with the severity of fall in single-kidney RBF and serum injury biomarkers in the draining renal vein [32, 35].

The ability of BOLD-MRI to monitor therapeutic interventions has been shown in groups of patients with RAS receiving autologous mesenchymal stem cells into stenotic kidneys [36] or pretreated with a mitochondrial targeted peptide before undergoing renal revascularization [37]. In both studies, a fall in $R2^*$ 3 months after treatment was associated with a modest increase in cortical perfusion. It should be emphasized that physiological changes in renal work may limit the ability to detect changes in $R2^*$. For example, in subjects with high-grade renal artery occlusion cortical and medullary $R2^*$ levels may appear normal due to complete lack of glomerular filtration and solute transport [38]. Some authors therefore propose interpretation of BOLD measurements in correlation with parenchymal renal volume and stenotic kidney GFR (measured by isotope renography) to derive a biomarker that may predict a positive renal functional response to revascularization [39].

Taken together, severe RAS may be characterized by either marked renal hypoxia consistent with tissue injury, and/or abolished response to furosemide, suggesting minimal residual blood flow to the kidney. BOLD-MRI may be able to assist clinical evaluation of RAS, but more data are needed to achieve clinically relevant advances.

Transplantation

In 2005, Sadowski et al. showed the feasibility of using BOLD-MRI in acute renal transplant rejection [40]. These authors found lower medullary $R2^*$ values in the acute rejection group compared with normally functioning transplants and transplants with acute tubular necrosis (ATN), while cortical $R2^*$ values were higher in ATN compared with the other groups. These observations were confirmed by several subsequent studies and the medullary-to-cortical $R2^*$ ratio has since been proposed as a marker to distinguish between these entities [41, 42]. The technique has been proven to be reproducible in subjects with good allograft function and its highest value appears to lie in differentiating stable well-functioning allografts from acute rejection, especially when decreased $T2^*$ values are measured in both the medulla and cortex [43]. BOLD-MRI has also been combined with perfusion and diffusion-weighted MRI to better understand the natural evolution of transplanted kidneys, their degree of fibrosis as well as the episodes of rejection/ATN that may occur when patients are followed over time [44].

Drugs and diet

BOLD-MRI is well suited for monitoring changes following acute dietary and/or pharmacologic manipulation. There are numerous reports evaluating different drugs or diets and the supplementary tables provide representative examples. Below is a brief discussion summarizing drug studies to date.

As mentioned, intravenous furosemide leads to an acute decrease in medullary $R2^*$ values and the response to furosemide is currently used as a functional parameter in studies with CKD [16] and RAS [38].

Drugs that modulate the renin-angiotensin system are commonly used by renal patients and have been shown to decrease $R2^*$ values in most but not all studies. Patients taking beta blockers showed lower $R2^*$ values compared with those that do not [45]. Inversely, increases in $R2^*$ have been observed after the infusion of angiotensin II [46]. In addition, one could observe indirect effects of cyclooxygenase inhibitors resulting in blunted response to water-loading in healthy young individuals [14].

Overall, there is good experience to date in evaluating effects of several drugs and diets with renal BOLD-MRI. Acute drug effects on renal oxygenation may provide early insights in the
In conclusion, BOLD-MRI has come a long way and is almost ready for prime time; international collaborations such as the ongoing COST action PARECHIMA offer unique opportunities to accelerate its entrance into clinical practice, most likely in combination with other functional MRI modalities within a single MRI session.

**SUPPLEMENTARY DATA**

Supplementary data are available at ndt online.

**ACKNOWLEDGEMENTS**

This article is based upon work from COST action 'Magnetic Resonance Imaging Biomarkers for Chronic Kidney Disease' (PARECHIMA), funded by COST (European Cooperation in Science and Technology). www.cost.eu. For additional information, please visit the PARECHIMA project website: www.renalMRI.org.

**FUNDING**

M.P. is supported by a grant from the Swiss National Science Foundation (FN 320030-169191), and P.V.P. by a grant from the National Institutes of Health (R01DK093793).

**CONFLICT OF INTEREST STATEMENT**

None declared.

**REFERENCES**


Received: 25.4.2018; Editorial decision: 27.6.2018