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Comparison of Magnetic Resonance Imaging With Radionuclide Methods of Evaluating the Kidney

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Nuclear medicine and MRI provide information about renal perfusion, function (glomerular filtration rate), and drainage. Some tracers that are used in nuclear medicine (technetium-diethylene triamine pentaacetic acid ([^{99m}Tc-DTPA] and ^{51}chromium-EDTA) and some contrast media (CM) that are used for MRI (gadolinium-DTPA for instance) share the same pharmacokinetic properties, though, detection techniques are different (low-spatial resolution 2-dimensional projection with a good concentration-to-signal linearity for nuclear medicine and high-resolution 3-dimensional localization with nonlinear behavior for MRI). Thus, though based on the same principles, the methods are not the same and they provide somewhat different information. Many MRI perfusion studies have been conducted; some of them were compared with nuclear medicine with no good agreement. Phase contrast can reliably assess global renal blood flow but not perfusion at a tissular level. Arterial spin labeling has not proven to be a reliable tool to measure renal perfusion. Techniques using CM theoretically can assess perfusion at the tissular level, but they have not proven to be precise. To assess renal function, many models have been proposed. Some MRI techniques using CM, both semiquantitative (Patlak) and quantitative, have shown ability to roughly assess relative function. Some quantitative methods (Annet's and Lee's methods) have even showed that they could roughly estimate absolute renal function, with better results than estimated glomerular filtration rate. Quantification of drainage has not been much studied using MRI.
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Quite early in the MRI development, it was shown to be a promising tool to study kidney, for instance being able to make out between cortex and medulla[1] or showing that concentration ability disappears in a model of acute tubular necrosis, whereas perfusion is preserved.[2,3] Obviously, MRI has advantages, such as many contrast opportunities, better spatial resolution, and lack of irradiation. However, it is more sensitive to motion, so it may be less appropriate for infants if one prefers avoiding sedation.

This paper focuses on functional information that can be assessed using both nuclear medicine and MRI after injecting a diagnostic agent, namely, perfusion, renal function (glomerular filtration rate [GFR]), and drainage. Morphological assessment, doubtless better given by MRI, is not addressed here. Also, in the last 20 years, other functional parameters were measured using MRI, which cannot be addressed using nuclear medicine, such as blood oxygenation using blood oxygen level-dependent (BOLD) technique,[4] microstructure by diffusion studies,[5] elastic properties using magnetic resonance elastography,[6] and sodium concentration imaging using sodium-23 MRI.[7]

Dynamic renal scintigraphy (DRS) shows the biodistribution of glomerular (technetium-diethylene triamine pentaacetic acid [^{99m}Tc-DTPA]) or tubular (^{99m}Tc-mercaptoacetyltriglycine (MAG3), ^{99m}Tc-ethylenedicysteine, or ^{123}I-hippuran) agents through 2-dimensional (2D) projections, with poor spatial resolution. The signal is directly proportional to the amount of tracer in the region of interest (ROI) (linearity). MRI shows the biodistribution of contrast agent, mostly gadolinium (Gd)-labeled glomerular agents in either 2D slices or 3-dimensional (3D) volumes. In nearly all studies, the longitudinal relaxation ("T1 effect" with signal enhancement) is used so these techniques are called dynamic contrast-enhanced MRI. In some others, usually with ultrasmall iron oxide particles, the signal decay induced by transversal relaxation is used (T2 or

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T2* effect). The relation between concentration and signal is not linear and not even monotonic. Basically, DRS shows global quantities (extensive parameters add together [eg, mass, amount, volume, flow, and energy]). They cannot be directly measured by sampling. Thus giving information on the whole kidney. Conversely, MRI shows local concentrations (intensive parameters average together [eg, temperature, pressure, concentration, perfusion, and density]). They can be measured by sampling, thus sampling the kidney. So even if measurements are based on the same principle (showing the biodistribution of a diagnostic agent), the techniques vary in detail. Each technique has its own advantages and drawbacks.

For DRS, adding all the counts within the ROI directly gives the total amount of tracer in the kidney, making the assessment of the whole kidney easier. The kidney is not sampled, as for MRI, and the tracer for every nephron contributes to the final signal. Moreover, the entire extracted tracer can be directly measured in this ROI as long as it has not left through the ureters, that is, for a much longer time span than for MRI when transit along the nephron, with water reabsorption, changes the concentration.

Most MRI techniques provide GFR per unit volume. The value should therefore be normalized by kidney volume, even if some authors chose to use a sampling approach without normalization.[8] A low-function kidney could be a small kidney with normal function for its size; in this case, a sampling technique (showing intensive parameter) would show that its function is normal. It could also be a normal-volume kidney with decreased function per unit volume; in that case, a sampling technique would show that its function is abnormal.

MRI has the clear advantage of resolution. Measurement of arterial input function (AIF) has been made possible to quite a precision using MRI. It is also a clear advantage for duplicated kidneys.[9] In return, MRI manual processing can be very time consuming (mostly segmentation).

However, the signal dependence on concentration is complex (increased using T1 effects and decreased using T2 effects at high concentration). Complex mixing of various concentrations within a single voxel is generally assumed to be under the fast exchange regime, which means that all tissue is assumed to relax with a single, average, T1. However, fast exchange is probably not achieved between blood and tissue[10,11] and it is questionable between glomerules and distal tubules in the same voxel. To minimize the T2 effects and keep the relation as linear as possible, the amount of injected Gd is usually kept low, the optimal seems to be between 0.02 and 0.04 mmol/kg for some authors[12] and higher values of approximately 0.1 mmol/kg for others. To infer concentration from signal enhancement requires knowing this dependence. Many authors have used calibration phantoms for this. However, relaxivity is not the same in vitro and in vivo (both temperature and chemical environment may play a role).

Also, owing to ventilation, kidneys move during the acquisition so motion correction is recommended, lest significant errors be made. Of course, kidneys also move during DRS but this only induces signal mixing with the low-intensity background, which is probably well-enough corrected using background subtraction.

Perfusion

Nuclear Medicine

To assess renal blood flow (RBF), one of the gold standards is the clearance of para-aminohippuric acid (PAH). Considering that the extraction coefficient is high (90%) and quite constant, RBF can be expressed as

$$RBF = \frac{\text{PAH Clearance}}{0.9 \times (1 - \text{Hct})} \quad (1)$$

where *Hct* is the hematocrit value. RBF is an extensive parameter whereas perfusion (RBF divided by organ mass or volume) is intensive. Several indices were proposed to assess perfusion from DRS (Kirchner index,[13] Hilson index,[14] etc.), but perhaps the most physiological index in the Peters index, which is the ratio of RBF to cardiac output.[15-17] If the injected tracer acted like microspheres in kidneys, Peters index could just be calculated as the ratio of the plateau uptake, corrected for attenuation, to the injected activity. The idea by Peters was to consider that, in the first seconds during which no venous output occurs, the tracer acts like microspheres. To extrapolate the plateau, Peters considered that the microsphere-cumulated activity is proportional to the integrated AIF. In this approach, only the shape of AIF is needed, not its absolute value. This has been explained in detail elsewhere.[18]

Magnetic Resonanace Imaging

It was suggested very early that MRI could assess renal perfusion.[19] Many techniques can be used.[20] To assess perfusion some kind of labeling is required. In MRI, the following 3 families of techniques can be used: labeling by injecting magnetic contrast agents, physical magnetic labeling the longitudinal magnetization (arterial spin labeling [ASL]), and physical magnetic labeling the transverse magnetization (phase contrast [PC]). Some studies have assessed perfusion[21-23] qualitatively or semiquantitatively,[24] showing drug-induced variations, but most techniques are aimed at providing absolute values of perfusion.

Another attempt was also made to assess perfusion with intravoxel incoherent motion effect. The displacement of water in the capillaries takes place along many different directions, virtually randomly. Thus, at the macroscopic scale, it is similar to random diffusion and perfusion can be assessed with diffusion-sensitized techniques.[25] However, in practice, this smart technique shows a dependence on RBF but fails to assess it with a decent precision.[26,27]

Arterial Spin Labeling

The general principle of ASL is to destroy longitudinal magnetization inside the slice of interest. It then slowly grows again by longitudinal relaxation (T1 effect). Also, perfusion rapidly brings blood from outside the slice with full magnetization. This increases the signal and makes it possible to calculate perfusion at the tissular level. Many variants of this technique have been published. First attempts in the kidneys were made in 1994 in rats,[28] then in 1995 in humans, with values of 2.78 ± 0.55 mL/min/g in the cortex and

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