Evaluation of flow measurement from the first pass bolus T_1 weighted images using inversion recovery sequence

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Objective: Previous studies have shown that the organ blood flows (OBFs) calculated using the T_1 weighted MRI technique were lower than the expected values. The aim of this study was a flow measurement comparison between the theoretical and experimental flows based on the technique before and after corrections (coil non-uniformity and inflow) using a flow phantom at two different concentrations (0.8 and 1.2 mmol l⁻¹).

Methods: A flow phantom was designed to produce three different flow rates at the same time. Theoretical flow was calculated by measuring the volumes of the phantom and dividing them by the time taken to fill these volumes. T_1 weighted turbo fast low-angle shot images were used to measure signal intensity (SI) change during the first bolus passage of the contrast medium through the phantom using linear phase-encoding acquisition.

Results: The corrected experimental flow based on the technique shows a good agreement with the theoretical flow, where the flow rate is low at the two concentrations.

Conclusion: The T_1 weighted MRI technique after the two correction factors can be used to measure the absolute flow where the flow rate is low, such as in the capillaries. For measuring high flow rate (e.g. artery), additional correction factors should be considered.

Received 16 November 2009

Revised 3 April 2010

Accepted 20 April 2010

DOI: 10.1259/bjr/18588668

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Several methods have been used for determining the haemodynamic parameters of the brain, including transcranial Doppler ultrasonography (TCD), dynamic or xenon-enhanced CT (Xe-CT), single photon emission CT (SPECT), positron emission tomography (PET) CT and MRI [1–5]. There are two main approaches to quantify the flow from contrast-enhanced data obtained using MRI, and these are the non-deconvolution or summary parameters methods and the deconvolution methods [6].

Moody et al [7] have applied a T_1 weighted MRI technique based on the microsphere technique [8] to calculate tissue blood flow to MR cerebral perfusion scanning. Organ blood flow can be calculated by measuring the gradients of the tissue signal intensity curve as a bolus of contrast agent passes through it and then dividing that value by the peak value of the arterial input function (AIF) based on the T_1 weighted MRI technique (T_1 technique).

Moody et al state that measurement of the asymptomatic middle cerebral artery (MCA) territory grey matter (GM) gave an average CBF of 42.6 ml 100 g⁻¹/min. This value was slightly lower than the other quantitative techniques such as SPECT, PET or Xe-CT.

Absolute renal blood flow measurement was reported by Vallee et al [9]. They used the T_1 weighted technique to obtain the flow from MRI. They reported the values of the cortical and the medullary perfusion, which were lower than the expected values from CT and PET.

In addition, Montet et al [10] used the T_1 technique to calculate absolute renal perfusion in the rabbits in 2003.

The MRI-derived perfusion was systematically lower than the expected values.

Until now, all perfusion measurements from the T_1 weighted MRI technique are lower than the expected values.

Therefore, some corrections should be applied on the equations, which were used for measuring absolute organ blood flow from the T_1 technique.

Two common acquisition strategies in MRI are linear phase encoding and centre out phase encoding [11]. This study is based on linear phase-encoding acquisition by use of T_1 weighted turbo fast low-angle shot (turbo fast low angle shot, TurboFLASH) images with application of the non-uniformity of the coil and the inflow effect corrections to evaluate the T_1 technique.

Aims

The aim of this study is to compare the theoretical (actual) and experimental flows based on the T_1 technique, before and after applying the correction factors of the non-uniformity of the coil and inflow effect, to assess the accuracy of T_1 technique for measuring flow. The experiments were performed at two different concentrations of Gd DTPA (gadolinium diethylenetriaminepenta acetic acid; 0.8 and 1.2 mmol l⁻¹).

Methods and material

Phantom

A flow phantom was designed to obtain different absolute flow rates (Figures 1 and 2). The shape of the phantom is approximately cubic and it is made of

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Figure 1. Picture of MR flow phantom with two branches of tube and one branch of cylinder. The phantom is divided into two parts. The first part is used for calibrating the absolute (theoretical) flow, which in turn is divided into three sections to produce three flows in the ratio 1:2:4. The effective volumes of these three sections are $V1=272$, $V2=136$ and $V3=68 \text{ cm}^3$. The second part contains three branches that can be used to produce three different flow rates at the same time.

Perspex. Its length, width and height are 20, 18 and 20 cm, respectively. The phantom is divided into two parts. The first part is used for calibrating the absolute (theoretical) flow, which in turn is divided into three sections to produce three flows in the ratio 1:2:4. The effective volumes of these three sections are $V1=272$ ($2 \times 8 \times 17$), $V2=136$ ($2 \times 4 \times 17$) and $V3=68$ ($2 \times 2 \times 17$) cm^3 . The three flows can be calculated by measuring $V1$, $V2$ or $V3$ and dividing them by the time taken to fill these volumes. Taps were adjusted so that the time taken to fill $V1$, $V2$ and $V3$ was the same during the experiments.

The second part contains three branches that can be used to produce three different flow rates at the same time. Each branch may contain material to simulate tissue. Three simulated "tissues" were used: two Tygon tubes (tissue A and tissue B) and a cylinder (tissue C). Tygon tubing was chosen because it does not absorb hydrophilic GD chelates. The internal diameter of the Tygon tubing is 0.95 cm. Since the innermost pixels of the tubes will be used for the experiments, they were assumed to produce the laminar flow. The cylinder, which contained 33 thin tubes, creates a uniform flow. The internal and external diameters of thin tube are 0.15 and 0.25 cm, respectively. The internal diameter of the cylinder is 1.65 cm. The phantom was placed inside the clinical head and neck coil.

A trolley, made from wood and aluminium, was designed for carrying the phantom to the MRI scanner room. It has three moveable shelves. There are three tanks on the shelves, two of which are water reservoirs, one above the other. By changing the height of the shelves and performing tap adjustments in the phantom, different flows can be obtained without using a pump (Figure 3).

Inflow effect

Inflow effects result when flowing liquid enters the imaging slice. In this situation, the partially saturated

liquid remaining in the slice after the radio frequency (RF) excitation from the previous sequence is replaced by fresh, unsaturated liquid. The strong signal from unsaturated water reflects its full magnetisation [12]. The inflow effect of the T_1 weighted TurboFLASH images appears between consecutive FLASH excitations. Since the velocity, which is dependent on the flow rate, can affect the signal intensity (SI), this effect should be considered for measuring the flow using the T_1 technique.

For measuring the inflow effect, the phantom was adjusted to produce four different flow rates (1.37, 2.74, 5.47 and 9.58 ml s^{-1}) at the same time. The two tanks contain the same concentration of contrast agent for the steady-state flow and stationary state. The stationary state can be obtained from the steady-state flow when water flow is stopped. Flow rates were calibrated by measuring the flow volume during a calibration time (about 50 s). The velocity was calculated from the flow divided by the inner cross-sectional area of the tube. The velocity of this experiment was calculated from the flow of the object (arterial or tissue, see "Theoretical and experimental flow") divided by its area (0.71 cm^2). The velocities were 1.93, 3.85, 7.72 and 13.49 cm s^{-1} , which covered the velocity of capillaries and big vessels. These experiments were done at two concentrations of Gd-DTPA (0.8 and 1.2 mmol l^{-1}) similar to the flow measurements. Previous study has shown that at low concentrations ($<2.1 \text{ mmol l}^{-1}$) of Gd DTPA, a near linear relationship with SI is observed; however, signal intensity response became non-linear at high concentrations [13].

The correction factor for inflow can be calculated from the SI of the steady-state flow divided by the SI of the stationary state at the same position.

Since the velocity, which is dependent on the flow rate, can affect SI, this effect should be considered when measuring the experimental flow.

To correct the experimental flow, the arterial input SI and gradient of the tissue should be divided by these factors. If the arterial input and tissue contain the same flow rate, the inflow does not have any effect on the experimental flow, because the correction factors will cancel out. But if the flow rates of the arterial input and tissue are not the same, the correction factors for the both of them should be calculated and applied to the flow measurement.

Coil non-uniformity

It is impossible to compare pixel intensities either qualitatively or quantitatively if the coil is not uniform. One of the major sources of image non-uniformity in the MR scanners is the RF coil inhomogeneity [14, 15].

There are two common methods for correcting intensity in MRI [16, 17]. The first involves the generation of a correction matrix from images of a uniform flood phantom, which is named the prospective method. The main source of error in this method is that the image of any "flood" phantom is itself subject to noise and, therefore, extra-statistical noise is added to the correction matrix and hence to the image which is to be corrected for non-uniformity [18]. The second uses digital filtering

Figure 2. Schematic diagram of MR flow phantom with reservoirs and main tanks. Water pressure in the main tank is constant during the experiment so long as water is constantly flowing in the overflow tube. Volumes V1, V2 and V3 are used to calibrate the absolute flow. Flow can be calculated by measuring V1, V2 or V3 and dividing them by the time taken to fill these volumes. The bulk of the phantom is filled with tap water or a saline solution.

(retrospective method) to remove the low-frequency components, which contain the non-uniformity. Using the retrospective method the non-uniformity is removed by filtering the images with a low-pass filter and

dividing the original image by the resulting low-frequency terms to remove the non-uniformity. It can be applied to any MR image, since it only uses the information naturally occurring in an image. In many cases this correction is implemented slice by slice, meaning that intensities in one slice are not comparable with intensities in the next. Several investigators such as Lufkin et al [19] and Linkar et al [16] have been working on developing a correction scheme for image non-uniformity correction using the retrospective method. Retrospective methods cannot distinguish between variations induced by scanner properties and those resulting from patient-related sources [16]. The method of correction that was used in this study is similar to the uniform flood phantom or prospective method.

For measuring the flow with the T_1 technique (see below) and inflow effect, the response of the RF coil should be uniform. The non-uniformity of the coil should be measured and applied to the SI for finding the corrected SI.

The non-uniformity of the coil was calculated from the SI of the stationary state with constant concentration. The SI of the centre of the phantom (e.g. tissue B) was chosen as reference SI. After normalisation of the mean SI from the stationary state, the correction factor can be calculated for different parts of the flow phantom. This correction factor is multiplied by the first pass bolus SI to give the corrected SI.

Figure 3. The figure shows the trolley for carrying the phantom to MRI scanner room, made with wood and aluminium. It has three moveable shelves. There are three tanks on the shelves, two of which are water reservoirs, one above the other. By changing the height of the shelves and performing tap adjustments in the phantom, different flows can be obtained without using a pump.

Theoretical and experimental flow

The theoretical flow for 1 pixel was calculated by measuring the volumes of the phantom (V1, V2 and V3, Figure 1) and dividing them by the time taken to fill these volumes as follows

$$\text{Theoretical flow for one pixel} = \frac{\text{measured volume}}{\text{time taken to fill the volume}} \times N_{\text{inside}} \quad (1)$$

where N_{inside} is the number of pixels inside the object being considered (*i.e.* tissue, organ, tube or cylinder). In this study, the experimental flow (based on T_1 technique) can be calculated by measuring the gradient of the tissue (*e.g.* the organ, tube or cylinder) SI time curve and then dividing that value by the peak value of the arterial input curve. The experimental flow of this experiment was measured from the mean SI of the 13 or 25 innermost pixels for the tissue A and tissue B or tissue C (cylinder), respectively:

$$\text{Experimental flow} = \frac{G_{\text{tissue}}}{SI_{\text{AI}}} \quad (2)$$

where G_{tissue} is maximum gradient of the tissue (organ or inside) and SI_{AI} is maximum signal intensity of the arterial input.

The non-uniformity of the coil is a major error for measuring and comparing the SI [14]. Since the experimental flow is based on the gradient of tissue curve and the amplitude of arterial input curve, the SI of these positions should be the true SI. Therefore, the correction factor of the non-uniformity of the coil should be multiplied by the gradient of the tissue and the amplitude of the arterial gamma-fit curve, to calculate the corrected experimental flow. Therefore, after non-uniformity correction, the experimental flow can be written as:

$$\text{Experimental flow (corrected coil)} = \frac{G_{\text{tissue}} \cdot cf_{\text{coil}}(\text{tissue})}{SI_{\text{AI}} \cdot cf_{\text{coil}}(\text{AI})} \quad (3)$$

where $cf_{\text{coil}}(\text{tissue})$ and $cf_{\text{coil}}(\text{AI})$ are the non-uniformity coil correction factors at the tissue and arterial input, respectively.

The velocity, which is dependent on the flow rate, can affect the SI, and as such this effect should be considered when measuring the experimental flow [20]. To correct the inflow effect on the experimental flow, the amplitude of both the arterial input and gradient of the tissue curve should be divided by these factors. Therefore, the experimental flow after correction of the coil non-uniformity and inflow effect can be written as:

$$\text{Experimental flow (corrected coil and inflow)} = \frac{\frac{G_{\text{tissue}}}{cf_{\text{inflow}}(\text{tissue})} \cdot cf_{\text{coil}}(\text{tissue})}{\frac{SI_{\text{AI}}}{cf_{\text{inflow}}(\text{AI})} \cdot cf_{\text{coil}}(\text{AI})} \quad (4)$$

where $cf_{\text{inflow}}(\text{tissue})$ and $cf_{\text{inflow}}(\text{AI})$ are the inflow correction factors of tissue and arterial input, respectively.

Figure 4. A coronal image of the phantom, which contains one cylinder and two tubes. Tissues A (highest flow), B (medium flow) and C (lowest flow, cylinder) contain three different flow rates. The position of the different assumed arterial inputs can be seen from this figure.

Figure 4 shows the coronal image of the phantom using the head and neck coil. It contains three different flow rates; tissue A (highest flow), B (medium flow) and C (lowest flow, cylinder), which contain 17.71, 17.71 and 35.5 pixels, respectively. To avoid the partial volume and laminar flow effect near the walls, 13, 13 and 25 innermost pixels were chosen for analysis of the tissues A, B and C, respectively. In addition, 13 of the 17.71 innermost pixels of the arterial inputs were also chosen for analysis to lessen the partial volume and laminar flow effect near walls. SI was calculated from the mean of the innermost pixels. Flow measurements may be made inside each object (simulating tissue or organ) using one of three input functions (arterial). Since each branch contains one simulated tissue and one, two or three arterial inputs, a range of different experimental flows can be calculated from various combinations (Figure 4). Since there is a limitation for using T_1 technique, just the main arterial input-2, named "arterial input", was used for calculation of the flow.

Since the total duration of the experiments is long for assessing the T_1 technique, both of the reservoir tanks were full of water.

It is necessary to mention that the experimental and theoretical values were measured in the MRI scanner room in the same position.

After perfusion image construction, the validity of the T_1 technical acquisition should be checked by the quality control (QC), which was one of the requirements of the model of Bell and Peters [21]. The QC ensures that the integrated arterial input curve multiplied by the tissue blood flow paralleled that of the brain tissue SI time curve [7].

Image acquisition

All studies were carried out using a 1.5T clinical MR scanner (Vision, Siemens Medical, Erlangen, Germany) using a standard head and neck coil. The phantom was

positioned centrally within the coil. Contrast was administered by a power CT injector, which was placed outside of the scanner room and was connected to the injection site in the phantom by an 8-m thin tube. T_1 weighted TurboFLASH images were used to measure SI change during the first bolus passage of the contrast medium through the phantom using linear phase-encoding acquisition. In addition, for calculating the non-uniformity of the coil, these image parameters were used to measure the SI of the steady flow and stationary state.

The acquisition parameters were echo time (TE) = 4 ms, time for one FLASH line = 8.5 ms, slice thickness = 10 mm, matrix size = 128×128 , inversion time set on scanner (TI) = 300 ms, effective TI = 844 ($300 + 8.5 \times 128 / 2$) ms, which is similar to null the signal from blood for linear phase-encoding acquisition [7], flip angle $\alpha = 15^\circ$. Images were acquired every 2 s.

Image analysis

The images data were transferred from the MR scanner to a UNIX workstation. The image processing software Interactive Data Language (IDL, Research Systems, Inc., <http://www.rsinc.com>) was used for processing.

Programs were written to automatically find the following.

1. The centre of gravity of each arterial input and tissue (inside or organ) and calculate the mean SI of the 13 innermost pixels of the arterial input (tube), tissue A and tissue B and the 25 innermost pixels for tissue C (cylinder) to avoid partial volume effects.
2. The correction factors of the non-uniformity of the different parts of the coil from the stationary state. The mean SI of the first pass bolus was then multiplied by these factors to find the corrected SI.
3. The inflow effect on the different flow rates.
4. The maximum amplitude of the arterial input and the maximum gradient of the simulated tissues (inside) from the gamma fit on the SI *vs* time curve. The program will be measured the experimental flow based on the T_1 technique, before and after correction of the non-uniformity of the coil and to compare with the theoretical flow (actual).
5. Finally to compare the corrected experimental flow (after corrections of the non-uniformity of the coil and inflow effect) with the theoretical flow. These programs could be run from either a UNIX workstation or a personal computer.

Results

Inflow effect

Figure 5 shows the effect of velocity on MR signal intensity at concentrations of 0.8 (red) and 1.2 mmol l^{-1} (black). The error bar shows the standard deviation of 13 innermost pixels. Figure 6 displays the relationship

Figure 5. Effect of velocity on MR signal strength, which was calculated at concentrations of 0.8 (red) and 1.2 mmol l^{-1} (black). The figure shows that an increase in the velocity is associated with increase in the signal intensity (SI). The error bar shows the standard deviation of 13 innermost pixels.

between inflow correction factor and velocity at the two concentrations.

As can be seen from the Figure 5 and Figure 6, an increase in the velocity and concentration is associated with an increase in SI and the inflow correction factor.

Coil non-uniformity

The mean non-uniformity coil correction factors were calculated as 1.86, 1, 1.05 and 1.98 for tissue C (after calculating the partial volume effect due to the thin tubes

Figure 6. Inflow correction factor, which was calculated from the steady flow over stationary state against velocity at concentrations of 0.8 (red) and 1.2 mmol l^{-1} (black). The figure shows that an increase in the velocity is associated with increase in the inflow correction factor. In addition, the correction value depends on concentration.

Table 1. Theoretical and experimental flows at concentration of 0.8 mmol l^{-1}

Flow (ml s^{-1})	Lowest flow (tissue C)	Medium flow (tissue B)	Highest flow (tissue A)
Experimental flow, without correction	0.04	0.23	0.40
Experimental flow, with the coil non-uniformity correction	0.02	0.11	0.21
Experimental flow, with the coil non-uniformity and inflow corrections	0.03	0.12	0.22
Theoretical flow	0.03	0.16	0.31

in the cylinder in a separate experiment), B, A and arterial input, respectively.

Experimental flow

The experimental flow was calculated using the T_1 technique at the two different concentrations. This will allow comparison between the three theoretical flows and the experimental flows at the two concentrations. The validity of the T_1 technique on the experimental flow was considered using QC [7]. The theoretical flow rates were measured as 0.03 , 0.16 and 0.31 ml s^{-1} using Equation 1. These flows were equivalent to velocities of 1.46 , 3.88 and 7.73 cm s^{-1} for a pixel (0.04 cm^2) in the tissues (C, B and A).

The experimental flow (Equation 2) was calculated from the mean SI of 13 or 25 innermost pixels for tissue A (highest flow) and tissue B (medium flow) or tissue C (lowest flow, cylinder), respectively (Figure 4). The two phenomena (non-uniformity of the coil and inflow effect) can affect the experimental flow. Therefore, when measuring the corrected experimental flow, the two correction factors should be applied.

The inflow correction factors for different velocities were obtained from Figure 6. The effect of inflow on the experimental flow was different for each tissue based on Equation 4. That is because the velocity of each tissue is different, requiring different inflow correction factors.

In summary, the theoretical flow was calculated based on Equation (1). The experimental flows before correction, after non-uniformity coil correction, and after the coil non-uniformity and inflow corrections were calculated from Equations (2), (3) and (4), respectively.

Tables 1 and 2 show the absolute values of theoretical and experimental flows at concentrations of 0.8 and 1.2 mmol l^{-1} , respectively.

The three different flows were calculated from tissue C (lowest flow, cylinder), tissue B (medium flow) and tissue A (highest flow) at a concentration of 0.8 mmol l^{-1} , as shown in Figure 7. Black (d) shows the ideal result. Red (e), blue (f) and magenta (g) show the experimental

flow before correction, after correction of the non-uniformity of the coil and after both corrections factor of the inflow effect and non-uniformity of the coil, respectively.

The theoretical flow *vs* corrected and uncorrected experimental flows at concentration of 1.2 mmol l^{-1} is shown in Figure 8.

In brief, the results indicate that after applying the correction factors, the T_1 technique can be used to measure the absolute flow where the flow rate is low, such as in capillaries.

Discussion

Peters et al [22, 23] have described a method for calculating blood flow from first-pass radionuclide studies based on the microsphere technique [8]. It is the gold standard for blood flow measurements in animal studies [7]. This technique is mathematically simple, requires no complex numerical analysis such as deconvolution and can be added to routine isotope renography. They used the technique to measure blood flow in the spleen and brain [21–23].

Miles [24] used this technique in dynamic CT for quantifying tissue perfusion in 1990. This was the first time that absolute values for tissue perfusion (kidney) from CT had been reported. This method is also used for measuring blood flow in a range of organs, such as the pancreas and brain, in CT [25–27].

Moody et al [7] have applied the Peters et al's [8] technique for calculating tissue blood flow to MR cerebral perfusion scanning on T_1 weighted imaging (the T_1 technique). This value was slightly lower than the other quantitative techniques.

Vallee et al [9] used the T_1 technique for measuring the absolute renal blood flow. The flow values were lower than the expected values from CT and PET.

Absolute renal perfusion measurement was reported by Montet et al [10] in rabbits by use of the T_1 technique. The results were also systematically lower than the expected values.

Table 2. Theoretical and experimental flows at concentration of 1.2 mmol l^{-1}

Flow (ml s^{-1})	Lowest flow (tissue C)	Medium flow (tissue B)	Highest flow (tissue A)
Experimental flow, without correction	0.03	0.03	0.55
Experimental flow, with the coil non-uniformity correction	0.02	0.13	0.29
Experimental flow, with the coil non-uniformity and inflow corrections	0.02	0.14	0.30
Theoretical flow	0.03	0.16	0.31

Figure 7. Theoretical against experimental flow at concentration of 0.8 mmol l^{-1} . Flow measurement from Equation (1) named "theoretical flow". Experimental flow measurement without correction named "experimental". Experimental flow measurement with the coil non-uniformity and the inflow corrections named "experimental (corrected)". The three different flows were calculated from the tissue C (lowest flow, cylinder), tissue B (medium flow) and tissue A (highest flow). Black colour (d) shows the ideal result. Red (e), blue (f) and magenta (g) colours show the experimental flow before correction, after correction of the non-uniformity of the coil and after both corrections factor of the inflow effect and non-uniformity of the coil, respectively.

The accuracy of the T_1 technique for measuring the flow after the non-uniformity of the coil and inflow correction was investigated in this study at concentrations of 0.8 and 1.2 mmol l^{-1} .

According to the results (see Figures 7 and 8) the following can be concluded.

1. The corrected experimental flow shows a good agreement with the theoretical flow at concentrations of 0.8 and 1.2 mmol l^{-1} for tissue C (lower flow, cylinder). The differences between the theoretical flow and experimental flow were 0.7% and 20.4% with concentrations of 0.8 and 1.2 mmol l^{-1} , respectively. Since there are enough data points on the tissue C curve, the maximum gradient can be calculated precisely and leads to accurate flow measurement. The slight difference may be due to the bolus dispersion (or an increase in mean transit time, MTT) between the arterial input and the tissue, because dispersion will decrease the maximum gradient of the tissue curve and lead to a decrease in the flow measurement. The difference between theoretical and experimental flow at concentration of 1.2 mmol l^{-1} also may be due to bolus dispersion or unknown error.
2. The corrected experimental flow in tissue B (medium flow) was in reasonable agreement with the theoretical flow at a concentration of 1.2 mmol l^{-1} . The difference between the theoretical flow and experimental flow was 7.9% at this concentration. The slight difference may be due to the bolus dispersion from the arterial input to the tissue, because, as mentioned above, dispersion can decrease the maximum gradient of the tissue curve and lead to a decrease in flow measurement. However, the corrected experimental

Figure 8. Theoretical against experimental flow at concentration 1.2 mmol l^{-1} . The three different flows were calculated from tissue C (lowest flow, cylinder), tissue B (medium flow) and tissue A (highest flow). Black (d) colour shows the ideal result. Red (e), blue (f) and magenta (g) colours show the experimental flow before correction, after correction of the non-uniformity of the coil, and after both corrections factor of the inflow effect and non-uniformity of the coil, respectively.

flow was lower than the theoretical flow at low concentration (0.8 mmol l^{-1}). The difference between the theoretical flow and experimental flow was 20.2% . Since the dispersion in tissue B should be the same for the two concentrations, the greater difference between the theoretical flow and experimental flow at low concentration (0.8 mmol l^{-1}) may be due to noise effect on the data.

3. The corrected experimental flows are lower than the theoretical flow for the tissue A (highest flow). The difference between the theoretical flow (0.3090 or 0.31 ml s^{-1}) and experimental flow (0.2177 or 0.22 ml s^{-1} at concentrations of 0.8 and 0.2959 , or 0.30 ml s^{-1} at a concentration of 1.2) was 30.6% and 6.4% at concentrations of 0.8 and 1.2 mmol l^{-1} , respectively. There is an increasing dispersion at 0.8 mmol . It seems that noise may have an effect on the data at low concentration (0.8 mmol l^{-1}). The experimental flow was just comparable with the theoretical flow at a concentration of 1.2 mmol l^{-1} . These differences may be due to a limited number of data points on the curve for tissue A, because the flow rate was high, and it leads to inaccurate measurement of its maximum gradient [28]. Another reason may be due to the dispersion of the contrast agent between the arterial input and the tissue.

Conclusion

The T_1 technique is relatively robust and mathematically simple and requires no complex numerical analysis, such as deconvolution. This technique is highly applicable in the research and clinical setting for measuring organ blood flow.

In summary, the results indicate that after applying the correction factors, the T_1 technique can be used to measure the absolute flow where the flow rate is low, such as in the capillaries.

It seems that for measuring high flow rates (e.g. in arteries), some additional correction factors to Equation (2) should be considered, because of inaccurate measurement of the maximum gradient of the tissue (or organ).

Our previous study shows that the inflow correction can be applied to the clinical data for measuring absolute perfusion [29].

Acknowledgment

The author gratefully acknowledges the data acquisition and manuscript editing of Dr Paul S Morgan.

References

1. Kudo L, Terae S, Katoh C, Oka M, Shiga T, Tamaki N, et al. Quantitative cerebral blood flow measurement with dynamic perfusion CT using the vascular-pixel elimination method: Comparison with $H_2^{15}O$ positron emission tomography. *AJNR Am J Roentgenol* 2003;24:419–26.
2. Reith W, Heiland S, Erb G, Benner T, Forsting M, Sartor K. Dynamic contrast-enhanced T_2^* -weighted MRI in patients with cerebrovascular disease. *Neuroradiology* 1997;39:250–7.
3. Harris G, Lewis R, Satlin A, English CD, Scott TM, Yurgelun-Todd DA, et al. Dynamic susceptibility contrast MR imaging of regional cerebral blood volume in Alzheimer disease: a promising alternative to nuclear medicine. *AJNR Am J Roentgenol* 1998;19:1727–32.
4. Siewert B, Schlaug G, Edelman R, Warach S. Comparison of epistar and T_2^* -weighted gadolinium-enhanced perfusion imaging in patients with acute cerebral ischemia. *Neurology* 1997;48:673–9.
5. Dean B, Lee C, Kirshch J, Runge VM, Dempsey RM, Pettigrew LC. Cerebral hemodynamics and cerebral blood volume: MR assessment using gadolinium contrast agents and T_1 -weighted Turbo FLASH imaging. *AJNR Am J Roentgenol* 1992;13:39–48.
6. Calamante F, Thomas D, Pell G, Wiersma J, Turner R. Measuring cerebral blood flow using magnetic resonance imaging techniques. *J Cereb Blood Flow Metab* 1999;19:701–35.
7. Moody A, Martel A, Kenton A, Alder S, Horsfield MA, Delay G, et al. Contrast-reduced imaging of tissue concentration and arterial level (CRITICAL) for assessment of cerebral hemodynamics in acute stroke by magnetic resonance. *Invest Radiol* 2000;35:401–11.
8. Peters A, Gunasekera R, Henderson B, Henderson L, Brown J, Lavvender JP, et al. Noninvasive measurement of blood flow and extraction fraction. *Nucl Med Commun* 1987;8:832–7.
9. Vallee JP, Lazeyras F, Khan HG, Terrier F. Absolute renal blood flow quantification by dynamic MRI and Gd-DTPA. *Eur. Radiology* 2000;10:1245–52.
10. Montet X, Ivancevic MK, Belenger J, Jorge-Costa M, Pochon S, Pechere A, et al. Noninvasive Measurement of absolute renal perfusion by contrast medium-enhanced magnetic resonance imaging. *Invest Radiol* 2003;38:584–92.
11. Jivan A, Horsfield MA, Moody AR, Cherryman GR. Dynamic T_1 measurement using Snapshot-FLASH MRI. *J Magn Reson* 1997;127:65–72.
12. Bradley WG, Waluch V. Blood flow: magnetic resonance imaging. *Radiology* 1985;154:443–50.
13. Nazarpour M, Moody AR, Martel AL, Morgan PS, 20th Annual Scientific Meeting, 18–21 September 2003, ESMRMB Rotterdam, S243–244.
14. Simmons A, Tofts P, Barker G, Arridge S. A. Sources of intensity nonuniformity in spin echo images at 1.5 T. *Magn Reson Med* 1994;32:121–8.
15. Condon B, Patterson J, Wyper D, Jenkins A, Hadley D. M. Image non-uniformity in magnetic resonance imaging: its magnitude and methods for its correction. *Br J Radiol* 1987;60:83–7.
16. Linkar B, Viergever MA, Pernus F. Retrospective correction of MR intensity inhomogeneity by information minimization. *IEEE Trans Med Imaging* 2001;20:1398–410.
17. Mohamed FB, Vinitski S, Faro SH, Ortega HV, Enochs S. A simple method to improve image nonuniformity of brain MR images at the edges of a head coil. *Comput Assist Tomogr* 1999;23:1008–12.
18. Condon BR, Patterson J, Wyper D, Jenkins A, Hadley DM. Image non-uniformity in magnetic resonance imaging: its magnitude and methods for its correction. *Br J Radiol* 1987;60:83–7.
19. Lufkin RB, Sharpless T, Flannigan B, Hanafee W. Dynamic-range compression in surface-coil MRI. *AJR Am J Roentgenol* 1986;147:379–82.
20. Nazarpour M. The inflow effect of signal intensity on T_1 -weighted images using inversion recovery (IR) sequence for measuring perfusion with MRI. *European Society for Magnetic Resonance in Medicine and Biology* September 2004;17(Suppl 1):S310.
21. Bell S, Peters A. Measurement of blood flow from first-pass radionuclide angiography: influence of bolus volume. *Eur J Nucl Med* 1991;18:885–8.
22. Peters A, Brown J, Crossman D, Hemingway AP, Roddie ME, Sallison DJ. Noninvasive measurement of renal blood flow with Technetium-99m-DTPA in the evaluation of patients with suspected renovascular hypertension. *J Nucl Med* 1990;31:1980–5.
23. Peters A, Brown J, Hartnell G, Myers MJ, Haskell C, Lavender JP. Non-invasive measurement of renal blood flow with ^{99m}Tc DTPA; comparison with radiolabelled microspheres. *Cardiovasc Res* 1987;21:830–4.
24. Miles K. Measurement of tissue perfusion by dynamic computed tomography. *Br J Radiol* 1990;64:409–12.
25. Miles K, Hayball M, Dixon A. Measurement of human pancreatic perfusion using dynamic computed tomography with perfusion imaging. *Br J Radiol* 1995;68:471–5.
26. Koenig M, Kraus M, Theek C, Klotz E, Gehlen W, Heuser L. Quantitative assessment of the ischemic brain by means of perfusion-related parameters derived from perfusion CT. *Stroke* 2001;32:431–7.
27. Klotz E, Konig M. Perfusion measurement of the brain: using dynamic CT for the quantitative assessment of cerebral ischemic in acute stroke. *Eur J Radiol* 1999;30:170–84.
28. Miles K, McPherson S, Hayball M. Transient splenic inhomogeneity with contrast-enhanced CT: Mechanism and effect of liver disease. *Radiology* 1995;194:91–5.
29. Nazarpour M, Moody AR, Martel AL, Morgan PS. Perfusion measurement on normal subjects with inflow corrected T_1 technique on T_1 -weighted images using inversion recovery sequences. *Magn Reson Mat Phys Biol Med* 2006;19(Suppl 1):355–6.