The Direction of Flow and Phase-encoding Schemes Effects on Signal Intensity in T1-weighted Inversion Recovery TurboFLASH Images

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ABSTRACT

Background: It is needed to minimize the effect of flow direction on the desired area, such as arterial input function (AIF) in magnetic resonance imaging (MRI).

Objective: The current study aimed to investigate the effect of flow direction on different velocities (0–80.39 cm/s) for the strength of the signal intensity (SI) at the linear phase-encoding (LPE) and the center out phase-encoding (COPE) schemes and to recommend the best flow direction in a selected slice and scheme for absolute perfusion measurement by inversion recovery T1-weighted turbo fast low-angle shot (TurboFLASH) MR images.

Material and Methods: In this experimental study, the flow rates were measured using a flow phantom, and the signal intensity (SI) was measured at the two opposite flow directions in the Z-axis perpendicular to the coronal image at a concentration of 0.8 mmol/L of gadolinium-diethylenetriaminepantaacetic acid (Gd-DTPA) by using the LPE and COPE schemes.

Results: The increase in velocity along with the growth in SI and inflow affected the use of LPE and COPE acquisitions in both directions. The velocity of the arterial input function is needed to calculate the inflow correction factor by using two schemes in two opposite flow directions to investigate perfusion.

Conclusion: The COPE scheme was better than the LPE scheme in measuring perfusion since the velocity and direction of blood flow affect SI less.

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Keywords

Perfusion; Cerebrovascular Circulation; Signal-To-Noise Ratio; Inversion Recovery, Contrast Agent; Flow Measurement; Linear Phase-Encoding; Center Out Phase-Encoding

Introduction

In magnetic resonance imaging (MRI), the region of interest (ROI) should be affected by the exciting and rephasing pulses. However, both pulses are always received by the nuclei in the stationary state, the flowing nuclei, used for the excitation in the slice, may exit before rephasing in the steady state, which is called the time-of-flight (TOF) with different effects on SI based on the type of pulse sequence and the phase-encoding schemes [1]. In other words, the results of the inflow effect are affected by inserting liquid into the imaging slice. The remaining liquid, which is partially saturated, in the slice from the previous sequence after

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radio frequency (RF) excitation replaces fresh and unsaturated fluid since the intensive signal from unsaturated water reflects the total magnetization [1,2]. In gradient echo sequences where the excitation pulse is followed by gradient rephasing, a flowing liquid has a higher signal than a stationary liquid [1]. Therefore, the inflow effect appears in the inversion recovery (IR) T1-weighted TurboFlash images between consecutive flash excitations.

An average flow velocity (v), leading to the replacement of all spins within the repetition time (TR, time for a FLASH line), enhances the signal in a slice section of a thickness (s) [3].

$$v = \frac{s}{TR} \quad (1)$$

Moody et al. [4] firstly used the T1-weighted technique for perfusion measurement in MRI. Also, this method is used in nuclear medicine and CT to measure blood flow in various organs (Equation 2). The organ's blood flow (OBF) or perfusion could be obtained from the concentration-time curve after contrast agent injection in MRI as follows:

$$OBF = \frac{G_{tisuue}}{SI_{AI}} \quad (2)$$

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

The maximum SI of the arterial input function (AIF) and the maximum gradient of the tissue or organ are obtained using the concentration-time curve in the ROI.

For measurement of cerebral blood flow (CBF), G_{tissue} is the maximum gradient of the concentration-time curve from the middle cerebral artery in the brain, and SI_{AI} is the maximum SI of the concentration-time curve from the common carotid artery in the neck as AIF.

In addition, the velocity effect on flow measurement must be determined due to its direct effect on the SI [5].

The correction factor for the inflow in position A can be calculated as follows:

$$Correction\ factor\ at\ point\ A = \frac{SI\ of\ the\ steady-state\ flow\ at\ A}{SI\ of\ the\ stationary\ state\ at\ A} \quad (3)$$

The gradient of organ or tissue and the arterial input SI from the concentration-time curve must be divided by the correction factors for greater accuracy. The absolute OBF can be then calculated using Equation 4 [6]:

$$Absolute\ OBF = \frac{\frac{G_{tissue}}{cf_{inflow}(tissue)}}{\frac{SI_{AI}}{cf_{inflow}(AI)}} \quad (4)$$

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

The inflow correction factors for both capillaries of tissue or organ blood flow and artery blood flow need computing to calculate absolute OBF (Equation 4). However, the estimation of the inflow effect on SI for the AIF and capillary is challenging in an *in vivo* study [6]. The accurate selection of image acquisition and the flow direction in the ROI is vital to minimize the inflow effect on estimating the absolute OBF.

Linear phase-encoding (LPE) and center-out phase-encoding (COPE) are the most common strategies in MRI [2].

However, the LPE acquisition starts at the top line of K-space and scans through subsequent lines, the COPE acquisition starts at the center of the K-space lines. Thus, one line can be read in each TR interval [7,8].

Many parameters can affect the SI and the relationship between SI and contrast agent concentration, including image parameters, such as the inversion time (TI) [9,10], saturation time (TS) [9,11], echo time (TE) [12], repetition time (TR) [13], flip angle [14], phase-encoding scheme [8], and image sequences [15]. In addition, the choice of contrast agents, whether that be Gd-DTPA [11] or iron oxide nanoparticles [16], as well as the magnetic field strength [17], have also been shown to have a measurable effect. The maxi-

mum SI of the arterial input can affect perfusion measurement using the T1 technique. The effect of inflow on SI at low velocities (0-13.5 cm/sec) was investigated in the previous study [2]. Also, the inflow effect of the contrast agent can change the strength of SI [6,7].

Due to the difficulty of the inflow correction measurement in clinical studies, the current study aimed to recommend the best phase-encoding schemes and direction of flow with less effect on SI of AIF for perfusion measurement.

This study includes two parts: 1) the investigation of the flow direction effect on SI strength at the two opposite flow directions in the Z-axis perpendicular to the coronal image of the phantom at different velocities (0–81.65 cm/s), which covered the velocities of the different AIF (e.g., the velocity of common carotid with a velocity of 22.28 cm/s) [6] and 2) the effect of various phase-encoding schemes on SI strength for moving the contrast agent by using IR T1-weighted turbo fast low-angle shot (TurboFLASH) images also the two directions.

The previous study investigated the effect of flow on SI by the two phase-encoding schemes at low velocities (0-13.5 cm/sec) just for one direction [2]. Therefore, low velocities cannot cover the AIF velocity (e.g., 22.28 cm/s) for perfusion measurement.

Material and Methods

Phantom

In this experimental study, a cubic flow phantom constructed from Perspex was designed to calculate different absolute flow rates. The length, width, and height of the phantom were 20, 18, and 20 cm, respectively. The size of the flow phantom was similar to a human head to fit in a clinical head and neck coil for more accurate simulation.

The phantom was divided into two sections: a) the calibration of absolute flow with the effective volume (V) of 476 cm^3 (Figure 1); the accurate flow was calculated by the required time for filling this volume, i.e., the exact flow

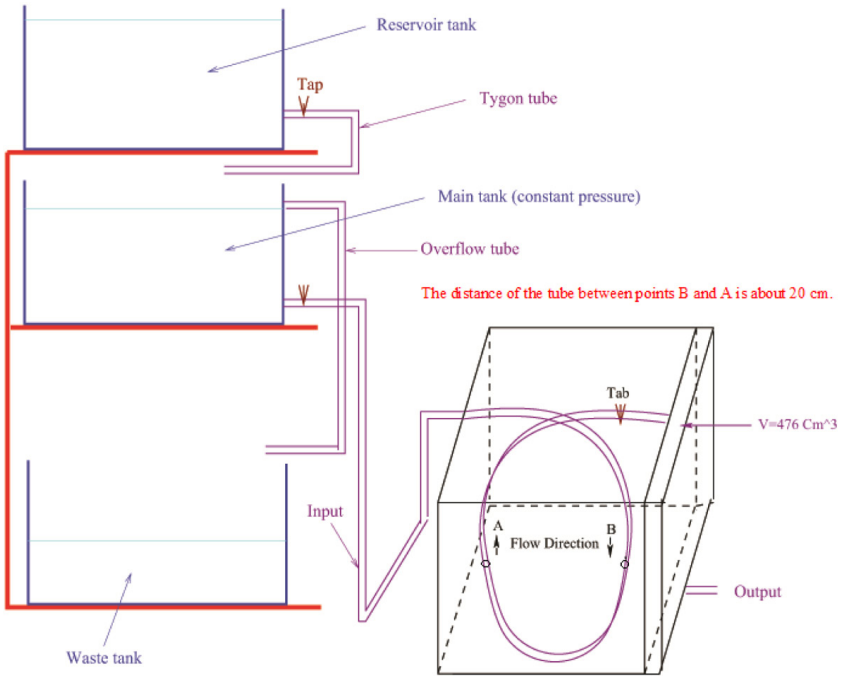

Figure 1: The reservoirs and main tanks of the magnetic resonance (MR) flow phantom that the two flow directions at the position of A and B are seen in the diagram with the 20 cm tube distance between points B and A.

is equal to the volume divided by the time and b) a branch of Tygon tubing that does not absorb hydrophilic Gadolinium (Gd) chelates [18]. The flow at the center of the tube was faster than that at the tube wall, in which the resistance of the tube wall slowed the flow, and tap water was inside the phantom. A plate fixed the tube with two holes, and different speeds can be created for AIF in different organs by adjusting the water valve.

In the phantom, the flow velocity changed from 0 to 80.39 cm/sec; for instance, in the common carotid artery, the velocity is 22.28 cm/s [19,20].

According to Smith et al. [21], the capillaries had a mean velocity of 0.02 to 0.03 cm/s.

The flow, which is equal to the volume of 476 cm^2 divided by the time this volume filled with liquid, was obtained by the pipe's cross-sectional area, which is 0.71 square centimeters, to measure the speed (Table 1).

Additionally, one tube was inside the phantom, connected between points A and B.

Therefore, the flow velocity is the same between points A and B, and the distance of the tube between the two points is about 20 cm (Figure 1).

The average of the nine innermost pixels was used for the SI calculation to eliminate the partial volume and laminar flow effects near the interior walls. The tube contained 17.71 pixels (2×2 mm^2). The steady state flow and stationary flow were simulated using the flow phantom. Furthermore, the stationary state could be generated while the water flow was stopped from a steady state flow.

A trolly with three moveable shelves made of wood and aluminum was used to carry the phantom to the MRI scanner room [5]. A continuous constant flow rate was obtained during the experiment by placing two constant-concentration reservoirs above the third tank. Changes in shelf height and phantom tap alignment could alter flow rates.

All three tanks used the same contrast agent concentration, 0.8 mmol/L of gadolinium-diethylenetriaminepantaacetic acid (Gd-DTPA, Magnevist, Schering Health Care Ltd, West Sussex, UK) for both the steady state flow and stationary state configurations.

Goyen et al. [22] reported the T1 relaxivity of Gd-DTPA (Magnevist 0.5 mol Gd/liter) in water (3.8 L/mmol. sec), 39 °C, 20 MHz, 1.5 T).

Figure 2 shows the coronal image of the phantom. The artifacts in the coronal images were possibly caused by the short read-out time of each Flash of the T1-weighted Turbo-Flash image or the movement of water inside the phantom.

Coil non-uniformity

In MR images, non-uniformity intensity can be made by the receiver filter, the gradient eddy current, a static magnetic field, gradient non-uniformity, size and shape of the image planes, and overall patient anatomy and position. Non-uniformity (e.g., RF coil inhomogeneity) is a parameter affecting SI in the MRI scanner [23]. The coil correction factor was calculated as in Equation 5 to eliminate this effect and obtain the corrected SI.

The normalized stationery in two opposite directions of the Z axis (A and B, Figure 1) perpendicular to the coronal image of the phantom are used as follows:

$$\text{Coil correction factor} = \frac{SI\ of\ the\ ROI\ (e.g.\ position\ of\ A)\ as\ references\ SI\ at\ stationary\ state\ flow}{SI\ for\ position\ of\ A\ or\ B\ at\ stationary\ state\ flow} \quad (5)$$

In the steady state for both flow directions (A and B), the corrected SI was calculated by

Table 1: Flow and velocity

Flow (mL/sec)	0.00	1.32	3.08	4.81	8.64	9.82	15.45	17.55	40.71	57.08
Velocity (m/sec)	0.00	1.87	4.34	6.77	11.91	13.84	21.76	24.71	57.34	80.39

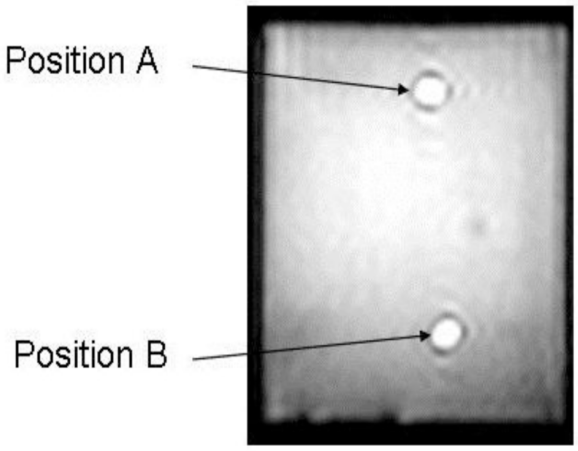

Figure 2: Coronal image of the phantom with positions A and B: the flow directions vertically inside and outside of the coronal image, respectively.

multiplying SI by its correction factor.

The inflow correction factor of these velocities at positions A or B was obtained as follows:

$$\text{Inflow correction factor} = \frac{\text{Corrected SI of the ROI (A or B) at steady state flow}}{\text{corrected SI of the ROI (A or B) at stationary state flow}} \quad (6)$$

Image acquisition

All studies were conducted on a 1.5 T (Vision, Siemens Medical, Erhlangen, Germany) clinical MR using a standard clinical head-and-neck coil. The phantom is placed inside and the center of the head and neck coil. T1-weighted TurboFLASH images were used to measure the SI of the stationary state and steady flow using LPE and COPE acquisition.

The TE and time to complete the FLASH line were set to 4 and 8.5 ms, respectively. The slice thickness was 10 mm, the matrix size was 128×128 mm^2, TR time was 2s, and pixel size was 2×2 mm^2. The flip angle was 15°, and TI set was 300 ms on scanner. Thus, the effective TI was 844 (300+8.5×128/2) ms, and the effective TI of 844 ms was similar to null point of the signal from the blood for the LPE acquisition. The COPE acquisition TI of 300 ms was used [4]; each image was acquired fifteen times.

Image analysis

The image processing software 'Interactive Data Language' (IDL version 8.5) was used for processing. The following were determined programmatically:

• The ten last acquisitions from 15 were averaged to improve the signal-to-noise ratio (SNR).

• The center of gravity in each ROI of the tube in both directions (A and B) to calculate the average SI of nine selected innermost pixels to eliminate partial volume effects.

• Universal network information exchange (UNIX) workstations or personal computers could also be used to run these programs. In this study, the phantom, trolly, and image parameters were similar to those in the previous work [2]; however, the velocity of the contrast agent, higher than that in the previous study, was investigated in the two directions.

Results

The mean non-uniformity coil correction factors were 1.00 and 1.15 using COPE acquisition and 1.00 and 1.10 using LPE acquisition for the A and B directions, respectively (Equation 4). Finally, these correction factors were applied to the mean SI of the nine innermost pixels to determine the corrected SI.

Figure 3 demonstrates the relationship between the velocity and corrected SI using the LPE and COPE acquisitions in the two flow directions (A and B).

Figure 4 illustrates the inflow effect versus velocity at the two positions for the LPE and COPE schemes calculated from Equation 6. The inflow correction factors for the high velocity (80.39 cm/s) were calculated as 1.48 and 1.29 for the LPE scheme, and 1.32 and 1.26 for the COPE scheme in A and B, respectively.

Discussion

The effect of flow on SI by the two phase-encoding schemes (COPE and LPE) at low velocities (0-13.5 cm/sec) just for one direc-

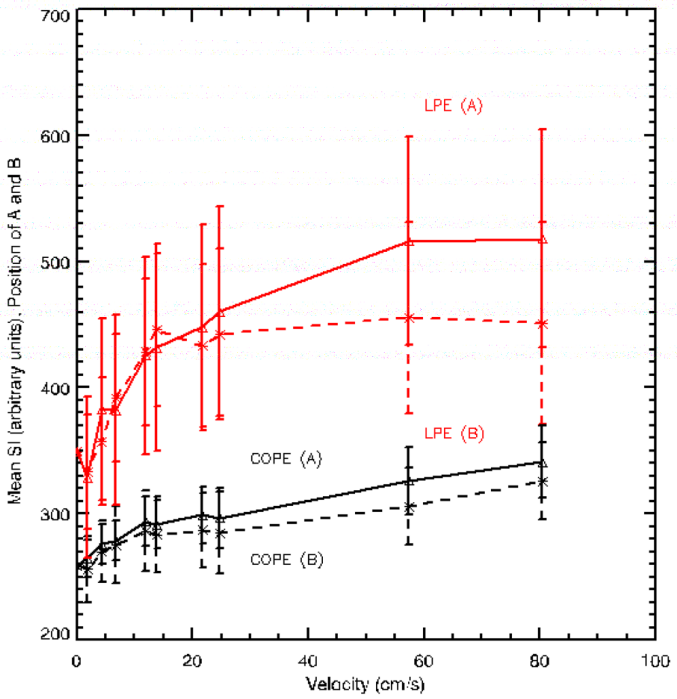

Figure 3: Effect of velocity on the strength of the magnetic resonance (MR) signal from the nine innermost pixels at the two positions for linear phase-encoding (LPE) (red) and center out phase-encoding (COPE) (black) schemes at the two positions (A and B). The standard deviation is shown by the error bar.

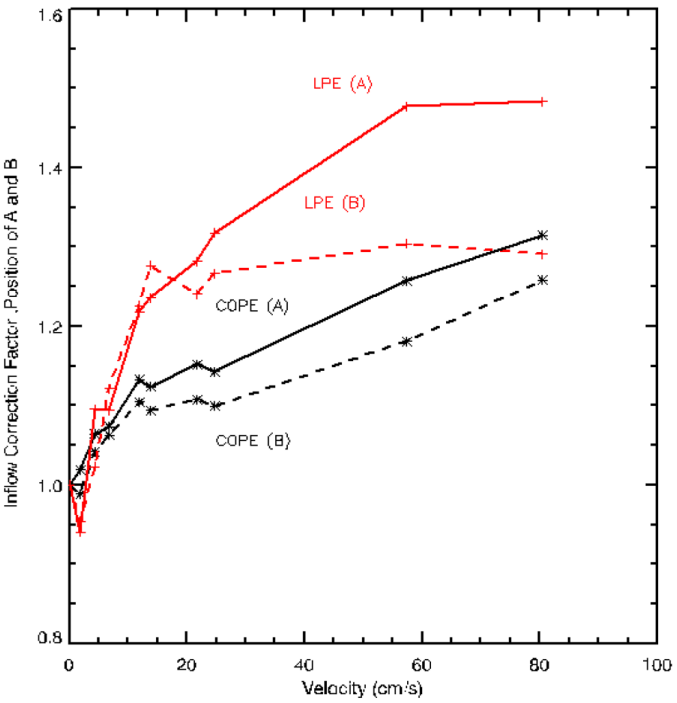

Figure 4: Inflow effect versus velocity on the strength of the magnetic resonance (MR) signal at the two positions (A and B) for linear phase-encoding (LPE) (red) and center out phase-encoding (COPE) (black) schemes.

tion was investigated [2]. Since these velocities could not cover the AIF velocity (e.g., 22.28 cm/s) for perfusion measurement [6], this study was conducted at different velocities (0–80.39 cm/s) with a concentration of 0.8 mmol/L of Gd-DTPA. The previous study showed that the relationship between SI and the contrast agent concentration is linear at a concentration of 0.8 mmol/L of Gd-DTPA [15]. These velocities covered the range of velocities in small vessels and the carotid arteries in humans.

The absolute OBF or perfusion is calculated after contrast agent injection (Equation 4) [24]. The previous studies indicated that the inflow effect of AIF was more critical than the inflow effect of the organ or tissue for perfusion measurement [6,7]. Additionally, in this study, the effect of the two flow directions (A and B) and phase-encoding schemes on the inflow effect was investigated to recommend the schemes with a lower velocity effect since the SI increases at a higher velocity.

Dean et al. [25] used a Sarns cardiovascular pump (Ann Arbor, MI) to produce a velocity between 0 and 70 cm/s of outdated blood at a concentration of 1 mmol/L of Gd-DTPAand also applied T1-weighted TurboFLASH acquisition (COPE) with image parameters of TI=529 ms, TE=4 ms. Also, they reported an inflow correction factor of 0.7 for a velocity of 20–40 cm/s to measure corrected cerebral blood volume.

Hacklander et al. [26] reported a correction factor of 0.9 for a blood velocity of 20 cm/s using T1-weighted imaging. Ivancevic et al. [27] developed a flow phantom to create different velocities (0-80 cm/s) with a concentration of 0.8 mmol/L of GD-DTPA and used a fast gradient recalled echo (LPE) sequence for perfusion measurement (effective TI=763 ms, TE=2.1 ms) with approximate correction factors 1.05 and 2.18 at velocities of 20 and 80 cm/s, respectively, used for the AIF for perfusion measurements.

The current study showed that the inflow correction factors were 1.28 and 1.48 for the position of A and 1.24 and 1.29 for the position of B at velocities of 20 and 80 cm/s, respectively, for the LPE scheme (Figure 4).

The difference between the results of this study and those of Dean et al. [25], Hacklander et al. [26], and Ivancevic et al. [27]'s studies is probably due to different image parameters and flow directions as they did not mention the direction of flow in the MR image [9,12,13,28].

Figure 4 shows that the inflow effects for a velocity of 24.71 cm/s (blood velocity of the common carotid artery, used as AIF for perfusion measurement [6]) were 1.32 and 1.14 at position A and 1.27 and 1.01 at position B for the LPE and COPE acquisitions, respectively.

The effect of low velocities (1.93–13.49 cm/s) on the SI using the two phase-encoding schemes was also investigated [2]. The inflow effects were 1.18 and 0.97 at a velocity of 13.49 cm/s for the LPE and COPE acquisitions, respectively (similar to the A direction) with a concentration of 0.8 mmol/L.

Based on the results, the inflow correction factors are 1.12 and 1.24 at a velocity of 13.14 cm/s for the COPE and LPE acquisitions, respectively, at A direction (Figure 4), showing the inflow correction factor in the LPE was higher than that in the COPE acquisition.

Peeters et al. [29] measured hepatic perfusion in nine patients using a saturated Turbo Field Echo (TFE) sequence and stated inflow caused an incorrect perfusion input function, resulting in a 30%-40% systematic error.

According to Peeters et al.'s study [29], a model designed based on the inflow is dependent on many factors, such as the contrast agent relaxation time, image parameters, and sequences. Therefore, the present study did not use a specific model for different image parameters and sequences.

Pang et al. [30] applied an additional prepolarized RF pulse to reduce the blood inflow effect for prostate tissue perfusion measurements using 3D T1W fast field echo (FFE).

Moreover, they concluded that an additional RF would significantly reduce the blood inflow effect, leading to less compromised T1 values and more accurate arterial input functions since a decrease in longitudinal relaxation times (T1) leads to an overestimation of AIF.

Based on the results of the current study, the slopes of the curves are not constant at low and high velocities (Figure 4). Therefore, it is impossible to identify a suitable linear equation model to fit the inflow correction factor at the two direction flows (A and B) for both LPE and COPE schemes. As some image parameters, such as TR, TE, FA, and TI and sequences affect the inflow correction factor, the identification of a specific nonlinear equation is considered a challenge for fitting the inflow correction factor at different velocities [29].

Moody et al. used a T1 MRI technique on T1-weighted imaging (LPE scheme) to measure the absolute CBF of 42.6 mL/100 g/min, which was lower than the SPECT CBF value [4].

The measurement of the absolute renal blood flow (the cortical and the medullary) with the T1-weighted technique, reported by Vallee et al. [31], was lower than the expected values from CT and Positron emission tomography (PET) Montet et al. utilized the T1 technique [32] for absolute renal perfusion in the rabbits and reported that the MRI perfusion was systematically lower than the expected values.

The T1 technique determined the CBF value by measuring the maximum tissue gradient and the arterial curve maximum amplitudes. Thus, the inflow effect of arterial input can lead to a low CBF.

Due to the similarity of image parameters and sequence of Moody et al.'s study [4] with this study, the inflow correction factor obtained from the LPE scheme was applied to clinical research. The inflow correction factor for a common carotid artery with a velocity of 22.28 cm/s is obtained at 1.23 (Figure 4). In addition, Figure 4 depicts that the inflow correction factor is approximately 1 for the capillary level (0.02–0.03 cm/s), ignored for measuring absolute CBF.

If this correction factor of 1.23 is applied on Moody's data, the corrected CBF value is computed 52.4 mL/100 g/min, which is consistent with the data obtained from SPECT (53.9 mL/100 g/min) [4].

Furthermore, the CBF value was 43.0 mL/100 g/min (standard deviation=8.8) on the middle cerebral artery territory gray matter of 11 normal subjects using the T1 technique [6]. Thus, the inflow correction factor (1.23) was applied to the clinical study without using a mathematical model as the image parameters, and sequences of the previous study were similar to those of the present study. Therefore, the corrected CBF was 52.9 mL/100 g/min, which in agreement with the PET literature [33-35].

In the clinical studies, the results of this test can be applied provided that the imaging parameters and sequence are similar to the current study.

Conclusion

In both acquisitions (LPE and COPE) the SI increased due to an increase in the velocity of the contrast agent. The inflow effect of the contrast agent on SI is related to its velocity within the ROI.

Therefore, the velocity of the contrast agent of the AIF is required to determine the inflow correction factor using the two schemes (COPE and LPE) for the two directions. Consequently, an organ's absolute blood flow or perfusion can be measured. The inflow effect measurement for AIF is difficult in a clinical study using LPE and COPE acquisitions. Because an increase in velocity correlated with increases in SI and the inflow effect.

Therefore, the COPE scheme in the direction of B had better acquisition than the LPE scheme for measuring perfusion since the velocity and direction of blood flow had less effect on SI.

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Ethical Approval

Not applicable, because this article does not contain any studies with human or animal subjects.

Conflict of Interest

None

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