Original Article

Inflow effect of Signal Intensity for the Centre out Phase-Encoding and Linear Phase-Encoding Acquisitions on Inversion Recovery T1-weighted TurboFLASH images

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Abstract

Background: *Two common acquisition strategies in MRI are Centre out Phase-Encoding (COPE) and Linear Phase-Encoding (LPE).*
Purpose: *The objectives of this study were to (1) to investigate the effect of concentrations (0.8 and 1.2mmol/L) and different velocities on signal intensity (SI) strength, (2) to find the inflow effect at different velocities at the two concentrations using the LPE and COPE acquisitions, and (3) to recommend the best acquisition for measuring perfusion with MRI using inversion recovery T1-weighted images.*
Methods: *A flow phantom was designed to produce four different steady state flows at the same time. The stationary state can be obtained when the water flow was stopped. All studies were carried out using a 1.5T clinical MR scanner. Mean SI was obtained in ROIs. The inflow correction factor can be calculated from the SI of the steady state flow divided by the SI of the stationary state at the same position in the image.*
Results: *The results show that an increase in concentration and the velocity is associated with increase in the inflow correction factor for the both acquisitions. But the inflow has less effect on SI with the COPE than the LPE acquisition. In addition, the correction value depends on concentration.*
Conclusion: *For measuring the absolute perfusion, the velocity and concentration of contrast agent of the arterial input function should be known to find the inflow correction factor using the two acquisitions. Otherwise, the COPE is the better acquisition for measuring perfusion, because inflow has less effect on SI.*
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Keywords: *Inflow effect • Inversion recovery • T1- weighted • Linear Phase-Encoding • Centre out Phase-Encoding*

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Introduction

To produce a signal with MRI, a nucleus must received a radio frequency (RF) excitation pulse. The resulting signal is dephased by the magnetic field gradient used to form an image and hence it must be rephased before it can be detected. If a nucleus receives the excitation pulse only and is not rephased, it does not result in a signal during the expecting data sampling time window. Similarity, if a nucleus is rephased but has not previously been excited, it does not produce a signal. Stationary state nuclei always receive both excitation and rephasing pulses, but in the steady state flowing nuclei present in the slice for the excitation may exit the slice before rephasing. This is called time of flight phenomenon and its effect depends on the type of pulse sequence used [1]. In the gradient echo (GE) pulse sequences the excitation pulse is followed by gradient rephasing. The RF pulse excites each selective slice but the rephasing gradient will be applied to the whole subject. Therefore, a nucleus in the flowing state that receives an excitation pulse is rephased regardless of its position and produce a signal. Stationary nuclei which receive repeated RF pulses are saturated because of a short TR (usually used with GE sequences). But when flowing nuclei enter the imaging slice, the partially saturated nuclei remaining in the slice after the RF excitation from the previous line of K space are replaced by fresh, unsaturated nuclei. The strong signal from unsaturated nuclei reflects their full magnetization. Therefore, the flowing liquid appears to have higher signal than stationary liquid. In summary, flow signal enhancement will occur in GE pulse sequences and these pulse sequences are often said to be flow sensitive [1-2].

Theoretical Background

Two common acquisition strategies in MRI are Linear Phase-Encoding (LPE) and Centre out Phase-Encoding (COPE) (see figure1). The COPE acquisition starts with the center of K space lines and moves out from there, jumping back and forth from positive to negative K space. The LPE acquisition starts from the top line of K space and scans through subsequent lines. Each line can be read after one TR interval [3].

Figure 1 shows a schematic representation of the magnetization during T1-weighted TurboFLASH sequence showing the ideal and real shape of the longitudinal magnetization recovery.

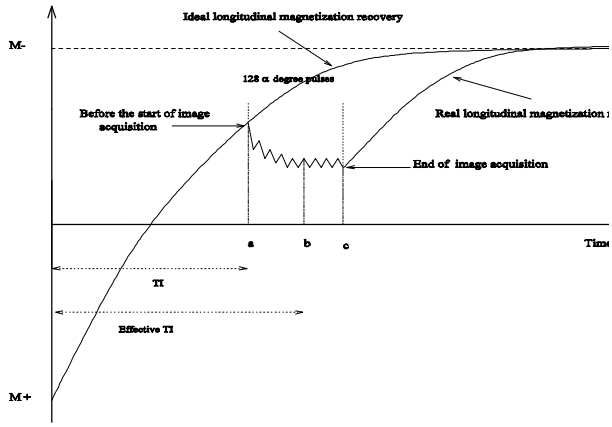

Fig 1- Schematic representation of the magnetization during T1-weighted TurboFLASH sequence showing the ideal and real shape of the longitudinal magnetization recovery. M- and M+ are the longitudinal magnetization values before and after the inversion pulse. After applying an inversion pulse, which occurs at time zero, as the longitudinal magnetization recovers, images can be acquired. Each image is formed by applying radio frequency (RF) (α = 15°) pulses, each of which gives one K space line. Point "a" shows the start of the RF pulses. The amplitude of signal after the first α is higher than the subsequent α. The time from the inversion pulse (TI) to the mid line of K space is usually called the effective TI. The effective TI (b) is placed at TI + 64×8.5 ms for a typical 128×128 matrix size image and for one FLASH line (8.5 ms).
The COPE acquires the centre of K space at point "a" and the LPE acquires the center of K space at point "b".

After an inversion pulse, which occurred at time zero, as the longitudinal magnetization recovers images can be acquired with TurboFLASH image sequence [4]. Each image can be formed by applying a set of α pulse, each of which gives one K space line The signal will decrease after each α pulse until a steady state is reached. After this step, it will be constant during image acquisition.

Figure 2 shows a schematic representation of the stationary state on T1-weighted TurboFLASH sequences. The α pulse causes increasing saturation through the sequences until all spins in the slice reach a steady state saturation. After this step the

amplitude of signal should be constant for the remaining α pulses[4].

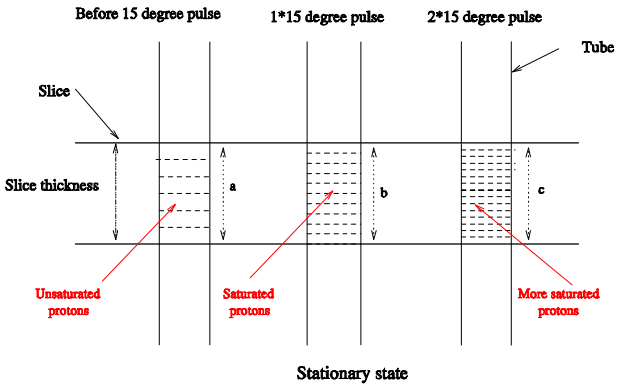

Fig 2- Schematic representation of the stationary state on T1-weighted TurboFLASH images. The time for one FLASH line is assumed to be 8.5 ms (each α = 15°). Before the initial α pulse the protons of liquid inside the slice are unsaturated (a) and emit a very strong signal. After the initial α the slice contains saturated protons (b) and gives a signal lower than before the initial α After the second α the slice contains more saturated protons (c) than after the first α and gives a signal weaker than from the first α. The unsaturated protons give a stronger signal than the saturated protons. In the figure, the higher density of "-" represents more saturation.

Figure 3 shows a schematic representation of the steady state flow on T1-weighted TurboFLASH sequences. The α pulse causes increasing saturation through the sequences. Flow enhancement will occur after each α.

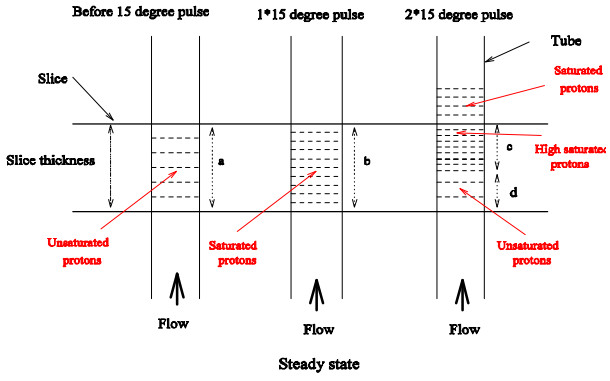

Fig 3- Schematic representation of the inflow effect on T1-weighted TurboFLASH images.

The time for one FLASH line is assumed to be 8.5 ms (each α = 15°). Before the initial α pulse the protons of liquid inside the slice are unsaturated (a) and emit a very strong signal.

After the initial α the slice contains slightly saturated protons (b). After the second α the slice contains a high number of saturated protons from the previous α (c) and a few unsaturated protons from fresh inflowing liquid (d), and gives a signal weaker than after the first α. The unsaturated protons give a stronger signal than the saturated protons. In the figure, the higher density of "-" represents more saturation.

A schematic representation of the magnetization for a steady state flow and stationary state of the longitudinal magnetization recovery (Mz) at a constant concentration is shown in figure 4. The inflow effect of the T1-weighted TurboFLASH images appears between consecutive FLASH excitations.

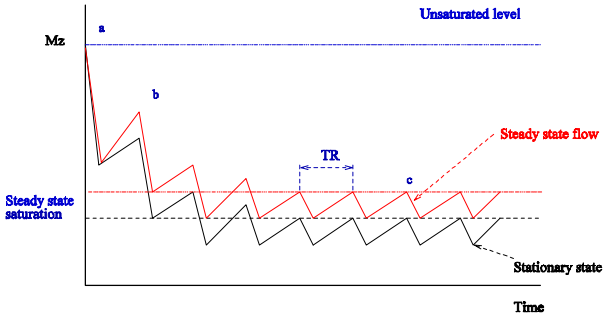

Fig 4- Schematic representation of the magnetization during T1-weighted TurboFLASH acquisitions for a steady state flow and stationary state of the longitudinal magnetization recovery (Mz). TR is the time between two α 's (8.5 ms). Flow enhancement will occur after each α in the steady state. Signal in the steady flow (red) is higher than the stationary state (black). After consecutive α, Mz will be reached at the steady state saturation. At this step, the strength of signal will not change for steady state flow and stationary state.

The "a" shows the unsaturated level of liquid before the initial α pulse. The COPE acquisition starts from the center of K space and the majority of image contrast is determined by the central views of K space and a small number of FLASH lines of K space around the centre. It will be affected by signal at point "a" and "b". Inflow and variation of T1, which is dependent on concentration, only has small effect on the SI in the COPE acquisition. The SI in the LPE acquisition is acquired at the centre of K space when Mz has reached steady state level (e.g. point "c") and hence exhibits greater variation with inflow.

As was mentioned before in detail (see figures 1, 3 and 4) the signal will decrease after some α to reach

a steady state saturation. After this step, it will be constant during the image acquisition. The COPE acquisition starts from the center of K space and the majority of image contrast is determined by the central views of K space and a small number of lines of K space around the centre. Inflow and variation of T1, which is dependent on concentration [5], only has small effect on the signal intensity (SI) in the COPE acquisition (see figure 4). The SI in the LPE acquisition acquires the centre of K space after 64 α 's (for 128 × 128 matrix size) when Mz has reached a steady state level and hence exhibits greater variation with inflow. The maximum signal enhancement can be expected when the average flow velocity (v) is such that all spins are replaced within the repetition time (time for one FLASH line TR) in a section of slice thickness (s) [6].

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

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.

Aims

The objectives of this study were (1) to investigate the effect of concentrations (0.8 and 1.2 mmol/L) and different velocities on SI strength, (2) to find the inflow effect at different velocities at the two concentrations by use of the LPE and COPE acquisitions, and (3) to recommend the best acquisition for measuring perfusion or flowing liquid with MRI using IR T1-weighted TurboFLASH images.

Methods

One of the parameters might be affected on SI is non uniformity in the MRI scanner. The uniformity gives an indication of how well the MRI system represents an object. One of the major sources of image non uniformity in the MR scanners is the radio frequency (RF) coil inhomogeneity [7-8]. For measuring SI 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 effect of velocity on MR corrected SI was measured using the LPE and the COPE acquisitions with a flow phantom. The non uniformity of the coil was calculated from the stationary state at different parts of the phantom and it was normalized to give the coil correction factor. To calculate the corrected SI for the steady state flow, SI of the each part of the phantom was multiplied by its correction factor, for the same concentration. In addition, the inflow correction factor of these velocities was calculated from the SI of the steady state flow over stationary state at the same position and at the same concentration. It is not necessarily to find the non uniformity coil correction factor for measuring the inflow correction factor, because the correction factor of the coil non uniformity is the same for the steady and stationary state at the same position and it will cancel out.

Phantom

A flow phantom was designed to obtain different absolute flow rates (see figure 5).

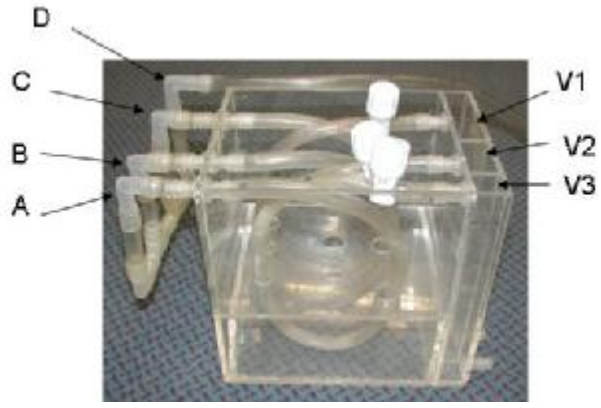

Fig 5- Picture of MR flow phantom. The phantom contains three branches. A, B, C and D contain four different flow rates. V1, V2 and V3 are the effective volumes for measuring different flow rates.

The size of the phantom is similar to the size of the head so that the coil (e.g head and neck coil) is presented with a similar dielectric load. The shape of the phantom is approximately cubic and it is made of 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 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 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 of Tygon tube. The internal diameter of the Tygon tubing is 0.95 cm. Tygon tubing was chosen because it does not absorb hydrophilic Gadolinium chelates (Gd) [9]. With add the main tube it can be used to produce four different flow rates; A, B, C and D at the same time. Flow rates were calibrated by measuring the flow volume during a calibration time. They were calculated as 1.37, 2.74, 5.47 and 9.58 mL/s for part A, B, C and D respectively. The tube contains 17.71 pixels (2×2 mm). To avoid the partial volume and laminar flow effect near the walls, 13 innermost pixels were chosen for analysis. SI was calculated from the mean of the innermost pixels. The phantom was used for creating a steady state flow and stationary state. The stationary state can be obtained from the steady state flow when the water flow was stopped. A trolley was designed for carrying the phantom to MRI scanner room, which was made with wood and aluminium. It has three moveable shelves. There are three tanks on the shelves, two of which are constant concentration reservoirs above the other to create a constant flow rate during the experiment. By changing the height of the shelves and performing tap adjustments in the phantom, different flows can be obtained without using a pump or siphon flow. The experiments were done at two concentrations of Gd-DTPA (0.8 and 1.2 mmol/L). Figure 6 shows the coronal image of the phantom.

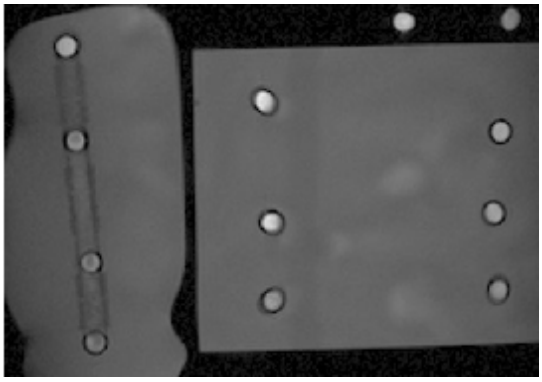

Fig 6- A coronal image with the COPE acquisition of the flow phantom using the head and neck coil

Image acquisition

All studies were carried out a 1.5 T clinical MR scanner (Vision, Siemens Medical, Erhlangen, Germany) using a standard head and neck coil. The phantom was positioned centrally within the coil. T1-weighted TurboFLASH images were used to measure the SI of the steady flow and stationary state using the LPE and COPE acquisitions. 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, repetition time [TR] = 2 s, pixel size 2×2 mm, flip angle [α] = 15°, inversion time set on scanner [TI] = 300 ms, effective TI = 844 (300 + 8.5 × 128/2) ms, which is similar to null the signal from blood for the LPE and COPE acquisitions. Images were acquired every 2 seconds. 15 acquisitions preformed for each image.

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 find automatically:

- The mean image of the 10 last acquisitions from 15 acquisitions, to improve the signal to noise ratio in the steady and stationary state flow.
- The centre of gravity of each region of interest for the branches (A, B, C and D) and calculate the mean SI of the 13 innermost pixels to avoid partial volume effects.
- To obtained the inflow effect on the different flow rates.

These programs could be run from either a UNIX workstation or a personal computer.

Results

The relationship between velocity and SI was investigated at concentration of 0.8 and 1.2 mmol/L using the LPE and COPE acquisitions.

LPE

Figure 7 shows the effect of velocity on MR signal intensity at concentrations of 0.8 (red) and 1.2 mmol/L (black) acquired with the LPE scheme. The figure shows that an increase in the velocity is associated with an increase in SI. The maximum signal enhancement in this experiment (time for one FLASH line = 8.5 ms, slice thickness = 10 mm) will occur at a velocity of 117.6 cm/s based on equation 1.

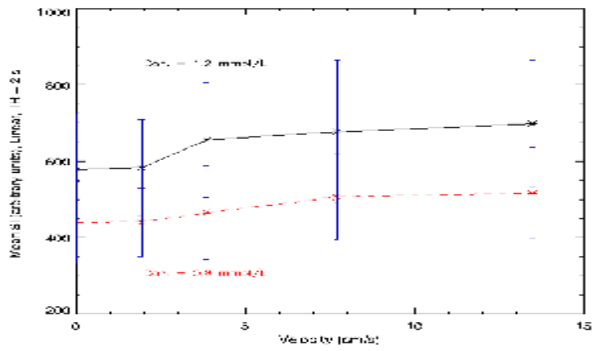

Fig 7- Effect of velocity on MR signal strength, which was calculated at concentrations of 0.8 (red) and 1.2 mmol/L (black) using the LPE acquisition. The figure shows that an increase in the velocity is associated with increase in SI. The error bar shows the standard deviation of SI for the 13 innermost pixels.

Figure 8 displays the relationship between inflow correction factor and velocity with the LPE acquisition at the two concentrations. As it can be seen from the figure, an increase in the velocity is associated with an increase in the inflow correction factor. In addition an increase in the concentration is associated with increase in the correction factor of the inflow effect.

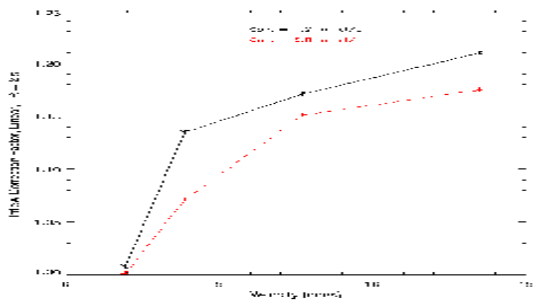

Fig 8- Inflow correction factor, which was calculated from the steady flow over stationary state against velocity at concentrations of 0.8 (red) and 1.2 mmol/L (black) using the LPE acquisition. The figure shows that an increase in the velocity is associated with increase in the inflow correction factor. In addition, the correction value depends on concentration.

COPE

Figure 9 shows the effect of velocity on MR signal intensity, which was calculated at the two concentrations (0.8 and 1.2 mmol/L) using the COPE acquisition. The figure shows that an increase in the velocity results in a small increase in SI. However, the range of SI spanned is much smaller than that observed using the LPE acquisition (see figure 7).

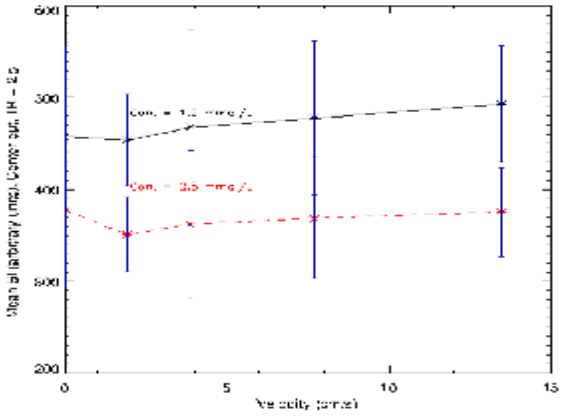

Fig 9- Effect of velocity on MR signal intensity, which was calculated at concentrations of 0.8 (red) and 1.2 mmol/L (black) using the COPE acquisition. The figure shows that an increase in the velocity results in a small increase in SI. The error bar shows the standard deviation of SI for the 13 innermost pixels.

Figure 10 depicts the relationship between inflow correction factor and velocity using the COPE acquisition at the two concentrations.

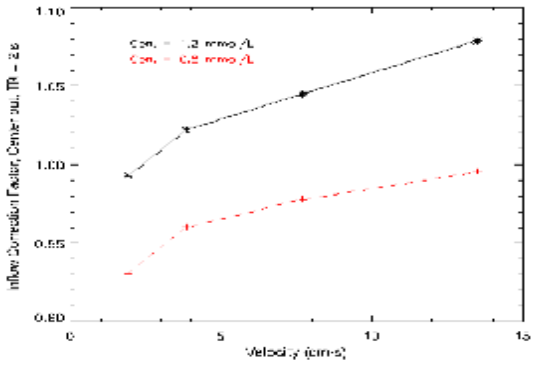

Fig 10- Inflow correction factor, which was calculated steady over stationary state against velocity at concentrations of 0.8 (red) and 1.2 mmol/L (black) using the COPE acquisition.

The figure also shows that an increase in the velocity results in a small increase in inflow

correction factors. However, the range of inflow correction factors spanned is much smaller than that observed using the LPE acquisition (see figure 8).

Discussion

Figure 7 shows that an increase in velocity is associated with an increase in SI using the LPE acquisition. As mentioned before (see figure 8) the LPE acquisition which is started from the top line of K space, is more sensitive to inflow compared with the COPE acquisition. The correction factor due to the inflow effect can be calculated from the SI of the steady state flow divided by the SI of the stationary state at the same position in the image. Figure 8 shows that the inflow correction factor is dependent on concentration and an increase in the concentration is associated with an increase in the inflow correction factor. According to the results, an increase in the velocity is associated with a large increase in the inflow correction factor (figure 8) on the LPE sequence. At a velocity of 13.5 cm/s (typical velocity of a big vessel in humans), the correction factor due to the inflow effect was 1.18 and 1.21 for concentration of 0.8 and 1.2 mmol/L respectively. The COPE scheme acquires the centre of K space before large variation of SI due to the inflow effect occurs (see figure 4), therefore the flow has much less effect using the COPE acquisition (see figure 9). As it can be seen from figure 9, SI in the stationary state is slightly more than the values obtained from the steady state at low concentration (0.8 mmol/L). This effect is not clear and may be due to poor slice profile from the short FLASH RF pulses. Figure 10 shows that the inflow correction factor slightly increased when the concentration increased from 0.8 to 1.2 mmol/L using the COPE acquisition. An increase in concentration leads to a decrease in T1, therefore faster recovery will occur in between the FLASH lines. Since the COPE acquisition starts from the center of K space and the majority of image contrast is determined by the central views of K space and a small number of FLASH lines of K space around the centre, it experiences a less of an increase in the inflow correction factor velocity. Flow related enhancement was reported by Dean et al. [10]. They pumped outdated blood with 1 mmol/L concentration of Gd-DTPA into a 9 mm (inner diameter) tube, which was positioned longitudinally through the scanner (Siemens 1.5 T). Image parameters from T1-weighted TurboFLASH acquisition were, time for one FLASH line = 8 ms, TE =4 ms, TI = 529 ms, slice thickness =10 mm. The velocity of blood was varied between 0 and 70 cm/s by a Sarns cardiovascular pump (Ann Arbor, MI). They calculated the inflow correction factor by dividing the SI of the stationary blood by the SI of flowing blood at the physiologic velocity of superior sagittal sinus. They demonstrated that, an increase in velocity up to about 30 cm/s is associated with an increase in SI. Dean et al. reported the flow related enhancement correction factor of 0.7 for superior sagittal sinus (assuming a velocity of 20-40 cm/s). This factor was multiplied by the SI to measure corrected cerebral blood volume in the superior sagittal sinus. Since Dean et al results are valid only for T1-weighted TurboFLASH sequence and its corresponding parameters; it is difficult to compare with our results. In addition, Hacklander et al [11]. Reported a flow related enhancement in the sagittal sinus from T1-weighted imaging. They assumed that the blood velocity is about 20 cm/s for the sagittal sinus. They empirically determined a correction factor of 0.9 for signal enhancement. This effect was taken into account during their data evaluation. Inflow effect concentration in fast gradient echo perfusion imaging reported by Montent et al. [12]. They made a phantom with a flow apparatus to determine SI as function of GD-DTPA concentration for various velocities. They show that an increase in concentration and velocities (0, 20, 30, 40, 50, and 80 cm/s) associated with increase SI using LPE acquisition. They used inflow correction to improve the accuracy of the arterial input function. Since they image parameter's (TR=5.74 ms, TE=2.1 ms, flip angel=90°, ...) are different than our image parameters and velocities, therefore they are impossible to compare with our results. Influence of the k-space trajectory on the dynamic T1-weighted signal in quantitative first-pass cardiac perfusion T1-weighted in MRI at 3T reported by Kim [13]. They compare the relative *T*1-weighted signal produced by the LPE and COPE *k*-space trajectories. They show that the COPE k-space trajectory yielded higher arterial input function (AIF) than the LPE k-space trajectories. The LPE k-space trajectory

yielded higher myocardial signal contrast (as a fraction of equilibrium magnetization) than the COPE k-space trajectories. They found the COPE acquisition is relatively optimal for the quantification of AIF than the LPE acquisition. In addition, they demonstrated that the SI for the LPE acquisition is higher than the COPE acquisition for the stationary state using a phantom containing 19 different concentration of Gd-DTPA. In spit of differences in strength of main magnetic fields (1.5 T vs. 3 T), the sequences (T1-weighted TurboFLASH vs. T1-weighted), and the image parameters between our study and Kim [13] study's, our results are similar to the myocardial and the stationary state of the Kim result's.

Conclusion

In conclusion, an increase in concentration and velocity associated with an increase in SI at the two acquisitions. The inflow effect depends on velocity and concentration. For measuring the absolute perfusion or organ blood flow [14-16], the velocity and concentration of contrast agent of the arterial input function should be known to find the inflow correction factor using the two acquisitions. Otherwise, COPE is the better acquisition for measuring perfusion, because inflow has less effect on SI.

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