motion; likewise, values obtained from the IS + IF − IF data were considered to be affected by IS motion. The estimated values were compared to those without MCs using a paired 2-tailed t test. P values <.05 were considered statistically significant. Data was expressed as mean ± 1SD.

MBF and PTF Estimation

Regional MBF and PTF were obtained using the single tissue compartment model with correction for partial volume effects and spillover from the left ventricular cavity (LV)

$$C(t) = \mathrm{MBF} \cdot \mathrm{PTF} \cdot C_a(t) \otimes \exp\left(-\frac{\mathrm{MBF}}{p}t\right) + V_a \cdot C_a(t)$$

where $C(t)$ represents the segment tissue time-activity curve (TTAC), $C_a(t)$ represents the arterial time-activity curve, p is the partition coefficient of water in the myocardial tissue (0.91 mL/g), and V_a is the arterial blood volume and spillover fraction from LV.[19] TTACs were generated for the nine myocardial segments using regions of interest (ROIs) selected within each ^{15}O-water myocardial image with and without MCs. The myocardial image was obtained by subtraction of the early phase of the dynamic ^{15}O-water image from the later phase. The midpoint between two phases was determined for each ^{15}O-water image using the contrast between myocardial regions and LV cavities. To obtain $C_a(t)$, a ROI was first drawn on a motion-corrected ^{15}O-CO image, without regard to whether MC had been applied to the ^{15}O-water image. The recovery coefficient of the LV was calculated from the ROI count on the ^{15}O-CO image and the blood radioactivity concentration. Second, the ROI was transformed to the ^{15}O-water image coordinate using a motion matrix that represents misalignment between transmission and the first dynamic frame of the ^{15}O-water scans. This transformation was made to evaluate motion effects on the ^{15}O-water scan, excluding effects of misalignment between ^{15}O-CO and ^{15}O-water scans. Using the transformed ROI, the LV TAC of the ^{15}O-water image was calculated. Finally, $C_a(t)$ was derived using the LV TAC and the overall TTAC was generated from nine myocardial segments according to the previous method.[6] To demonstrate the influence of IS motion, the ROI-based approach was not used; instead, pixel-by-pixel MBF and PTF values were estimated for a representative normal scan.[20] To denoise and smooth these parametric images, MBF and PTF were set to zero in voxels with PTF < 0.3 g/mL or V_a > 0.8 mL/mL, and then filtered using a Gaussian filter of 14 mm (full width at half maximum).[21] For normal scans in the group, improvement of homogeneity between myocardial segments by MC was also evaluated as $100 \times (1 - \mathrm{SD_W}/\mathrm{SD_{WO}})$, where $\mathrm{SD_W}$ and $\mathrm{SD_{WO}}$ were standard deviations of regional values with and without MCs, respectively.

RESULTS

Table 1 shows global movement during ^{15}O-water scans, categorized by the presence of IS and IF motions. Thirteen scans (7 scans on healthy volunteers and 6 scans on patients) exhibited IS motions greater than 4.0 mm, and 10 scans (5 scans on healthy volunteers and 5 scans on patients) exhibited IF motions greater than 4.0 mm. Among these, 4 scans on healthy volunteers and 5 scans on patients showed both IS and IF motions.

There was no statistically significant difference between the ROI volumes of ^{15}O-water images with and without MCs for IS, IF, and NE motion groups by paired 2-tailed t tests, and between the volumes with and without MCs, IS + IF − IS, and IS + IF − IF motion for IS + IF motion group by a one-way ANOVA.

Table 2 lists MBF and PTF values with and without MCs categorized by the four types of movement that occurred during ^{15}O-water studies, and average percent differences of MBF and PTF values over nine segments. Among 171 myocardial segments from all subjects, the fitting program failed to provide physiologically meaningful values in two segments (middle and basal septa for a healthy volunteer, in which IS + IF motion was detected) in the absence of MC. Those data were excluded from subsequent analysis. Among MBF values, significant changes by MCs were found in the IS + IF, IS + IF − IF, and IF motion groups: for IS + IF, from 0.845 ± 0.366 to 0.769 ± 0.319 mL/minute/g ($P < .05$); for IS + IF − IF, from 0.845 ± 0.366 to 0.780 ± 0.360 mL/minute/g ($P < .05$); and for IF, from 0.854 ± 0.179 to 1.088 ± 0.154 mL/minute/g ($P < .01$). PTF values in the IS motion group changed significantly from 0.465 ± 0.118 to 0.504 ± 0.087 g/mL ($P < .05$). This significant change is also shown in Figures 2 and 3, as differences between PTF values with and without MCs. In Figure 3, a data point indicated an averaged value for each scan.

Figure 4 illustrates the effect of IS motion of 5.0 mm on MBF and PTF of a normal subject from the IS motion group. The polar maps of normal PTF in Figure 4 demonstrate improvement of homogeneity in anterolateral regions by MC. Improvements of homogeneity of three normal scans in the group were 2.6% ± 26.1% for MBF and 30.9% ± 22.7% for PTF.

Table 1. Characteristics of motion during ^{15}O-water scans

Motion type	Total (normal) scans	IS motion (mm)	IF motion (mm)
IS + IF	9 (4)	7.7 ± 2.7	11.0 ± 4.0
IS	4 (3)	5.4 ± 1.0	2.8 ± 1.1
IF	1 (1)	2.7	5.1
NE	5 (4)	2.5 ± 0.7	1.4 ± 0.5

Table 2. The summary of MBF and PTF values with and without MCs

Motion type	Without MC	With MC	Percent difference
MBF (mL/minute/g)			
IS + IF	0.845 ± 0.366	0.769 ± 0.319*	33.8 ± 61.5
IS + IF − IS		0.815 ± 0.341	21.2 ± 38.5
IS + IF − IF		0.780 ± 0.360*	25.1 ± 41.4
IS	0.855 ± 0.343	0.828 ± 0.259	9.0 ± 8.9
IF	0.854 ± 0.179	1.088 ± 0.154†	30.2 ± 20.9
NE	0.898 ± 0.233	0.909 ± 0.221	9.7 ± 15.7
PTF (g/mL)			
IS + IF	0.476 ± 0.133	0.469 ± 0.128	13.9 ± 10.9
IS + IF − IS		0.471 ± 0.135	12.3 ± 13.0
IS + IF − IF		0.490 ± 0.119	12.3 ± 10.2
IS	0.465 ± 0.118	0.504 ± 0.087*	9.0 ± 9.6
IF	0.512 ± 0.067	0.483 ± 0.068	7.1 ± 5.6
NE	0.488 ± 0.095	0.496 ± 0.100	5.1 ± 4.9

IS, Inter-scan; *IF*, inter-frame.
NE denotes motion ≤4.0 mm. IS + IF − IS and IS + IF − IF represent IS + IF motion groups with correction of IS and IF motions, respectively.
* $P < .05$, † $P < .01$ vs without MC.

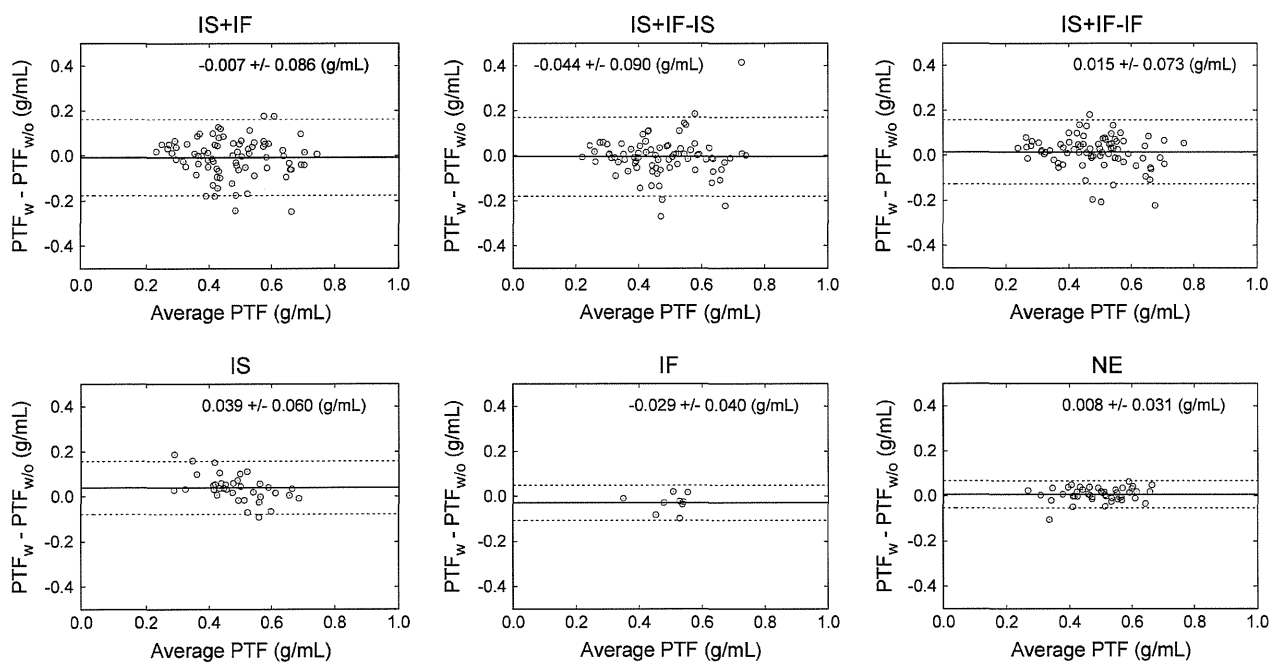

Figure 2. Differences between PTF values with and without the motion corrections. *IS*, Inter-scan motion; *IF*, inter-frame motion; *NE*, no significant motion. IS + IF − IS and IS + IF − IF are groups corrected for IS and IF motion in the IS + IF group, respectively.

Relative variability of regional MBF and PTF values due to global motion are shown in Figures 5 and 6, respectively. For MBF and PTF values in each motion group including the IS + IF − IS and IS + IF − IF groups, no significant difference between myocardial segments was observed in a one-way ANOVA (IF

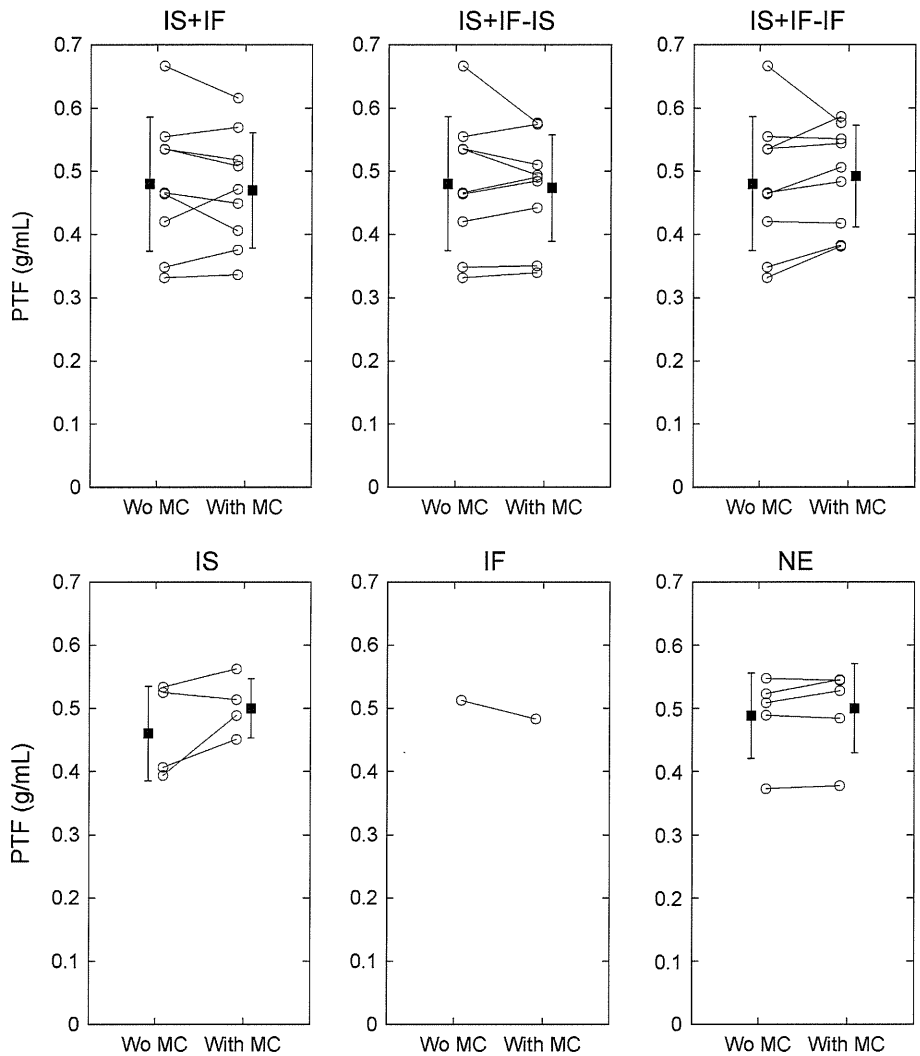

Figure 3. Changes in PTF values in the absolute scale by the MCs. *IS*, Inter-scan motion; *IF*, inter-frame motion; *NE*, no significant motion. IS + IF − IS and IS + IF − IF are groups corrected for IS and IF motion in the IS + IF group, respectively. A data point indicates an averaged value for each scan.

motion group was excluded from this analysis because one scan was assigned to this group. Data for IS + IF − IS and IS + IF − IF group were not shown).

DISCUSSION

In this study, global movements of the subjects during ^{15}O-water PET scans at rest were categorized as IS and IF motions. IS motion involves misalignment relative to the transmission scan, whereas IF motions involves changes between the dynamic frames during the ^{15}O-water scan. After categorization, the effects of these motions on MBF and PTF measurements were evaluated. Consequently, it was demonstrated that MBF values are affected by IF motion rather than IS motion.

Investigation of regional sensitivity to global movement resulted in no significant difference in regional MBF or PTF. This is because the direction and the magnitude of global movement varied for each subject; under- and over-estimation of MBF and PTF due to the motion could have occurred in any myocardial region. In this study, we utilized a one-way ANOVA to detect segmental differences. Due to small sample sizes, the ANOVA may fail to detect the differences. Correlation between segmental differences and the directions of global motions, especially IS motion, effects of which have been reported for misregistration between CT

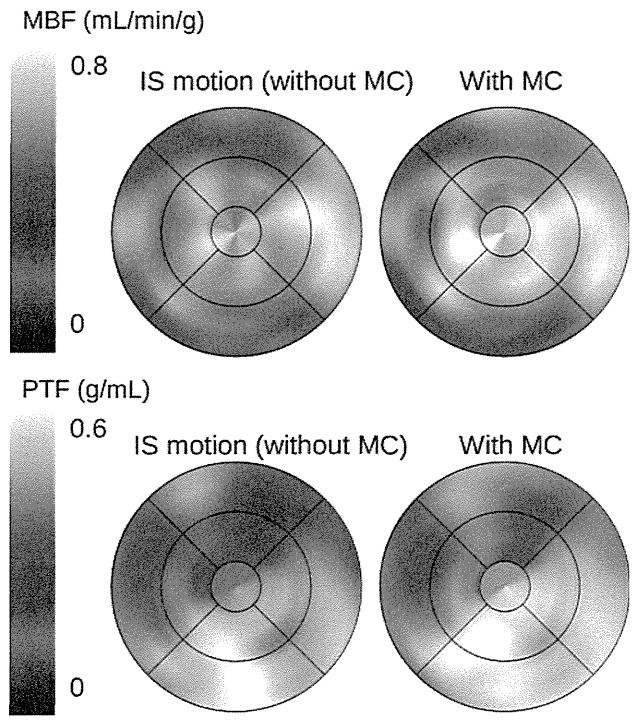

Figure 4. Effect of an IS motion of 5.0 mm on MBF and PTF of a normal subject in IS motion group.

attenuation map and emission image,[22,23] could be observed in large population study.

MBF values changed significantly upon correction for IF motions in the IF, IS + IF, and IS + IF − IF motion groups. In contrast, correction of IS motion was considered to have little effect, based on the results from the IS + IF − IS and IS motion groups in Table 2. It was further demonstrated that IF motion, but not IS motion, was an important source of error in the MBF measurement. The significant but small changes in MBF of IF, IS + IF, and IS + IF − IF motion groups after MCs could be due to the relatively large number of healthy volunteer scans in each group (1 of 1 scan for IF, 4 of 9 scans for IS + IF, respectively), as shown in Table 1. Suppression of IF motion during the ^{15}O-water scan is needed for accurate MBF measurement. For the IS motion groups, no significant change in MBF values and significant change in PTF values were observed by MCs (Figures 2, 3). For the normal scans in the group, the improvement of regional homogeneity was also observed in PTF rather than MBF. This is because the MBF values obtained using the kinetic model employed in this study were based on a clearance rate assessment of ^{15}O-water rather than the uptake rate of the radiotracer.[5] This is consistent with the findings of Lubberink et al,[21] who reported that MBF values do not change even if attenuation correction is omitted; this correction only caused changes in the absolute scale of TTAC, which then caused changes in PTF values. IS motions were also detected in the IS + IF motion group. For the IS + IF − IS motion group, in which IS motions were corrected, no significant change in PTF values was observed after MCs. This might be because IS and IF motions affected the ^{15}O-water images not subjected to MC in such a way as to cancel out errors in PTF measurements. PTF values were considered to be more affected than MBF values during the wash-in phase. However, in our study, IF motion had little effect on PTF values for the IS + IF − IF and IF groups, as shown in Table 2. The reasons for this discrepancy are follows: (1) to estimate the quantitative values, a nonlinear least squares method with same weights for all data points were applied to the data corrected for the physical decay. This fitting manner could be sensitive to radioactivity concentration in wash-out phase rather than wash-in phase. (2) Over-estimation in TTAC was introduced by contamination of the LV cavity count, using the ROI for TTAC superimposed on dynamic frames. However, the spillover correction could suppress the influence of IF motion. One of the advantages of ^{15}O-water over other cardiac PET tracers is that perfusion and viability measurements can be obtained from a single PET scan with short duration. Because accurate PTF measurement enabled assessment of

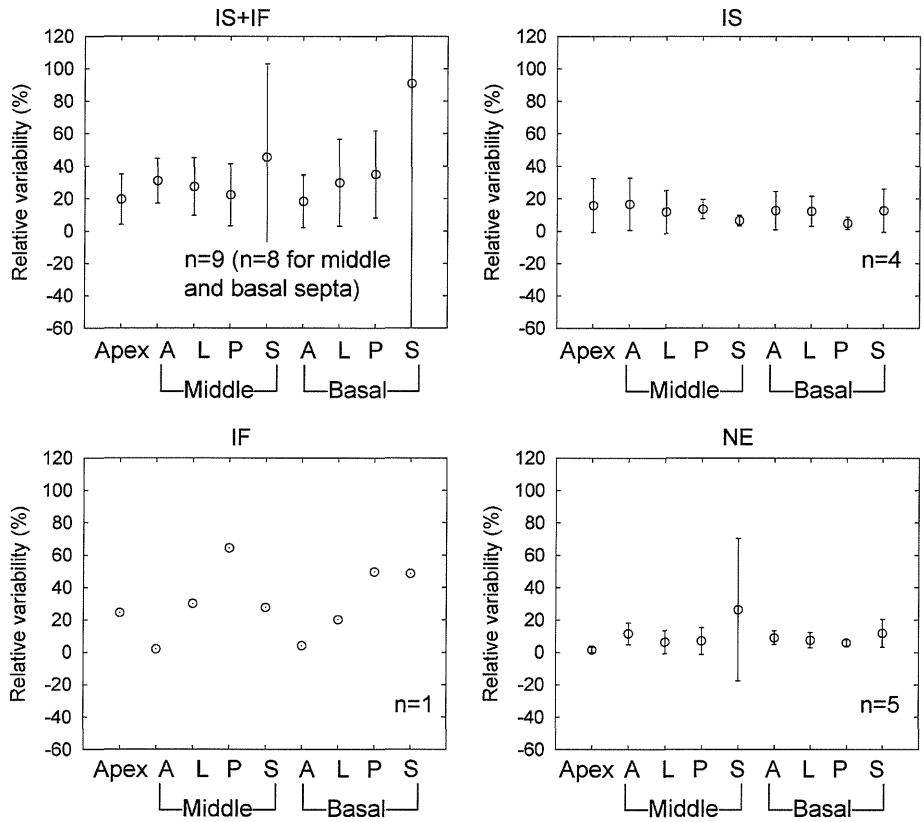

Figure 5. Relative variability of regional MBF due to effects of global motion. *IS*, Inter-scan motion; *IF*, inter-frame motion; NE represents no significant motion. IS + IF − IS and IS + IF − IF are groups corrected for IS and IF motion in the IS + IF group, respectively.

myocardial viability,[7-12] correction of IS motion, the effect of which was shown as the improvement of regional homogeneity of normal PTF in Figure 4, is considered to be important in the diagnosis of myocardial infarction, or evaluation of effects of cell transplantation therapies.

In this study, we performed ^{15}O-CO scans to determine a recovery coefficient for each subject.[5,6] ^{15}O-water PET studies without ^{15}O-CO scans have been reported in previous papers[3,21,24]; in these studies, the recovery coefficient could be fixed, or assumed to have a constant value. Omission of the ^{15}O-CO scan shortened the examination time, and might reduce motion artefacts. However, we considered that the adequacy of using a fixed recovery coefficient is still an unresolved issue in ^{15}O-water PET. We considered that the changes in MBF and PTF resulting from MCs were due to correction for global motion, and not due to variability of myocardial tissue ROIs drawn manually on ^{15}O-water images. We employed nine segmented regions rather than 17-segment AHA standard model for suppressing relatively large noise level of ^{15}O-water data.[4] In addition, the volumes of the ROIs for each motion group before and after MCs were almost equal (not statistically significant). Furthermore, lower sensitivity of MBF values to variation in ROI size and shape was demonstrated by Iida et al.[4] In this study, global motion might have occurred because subjects were scanned without any fixation. Although tight fixation prevents patient motion, such fixation could bring discomfort or pain to the patient; subsequent reaction to such pain could itself induce motion.

One limitation of this study is that all subjects were scanned at rest. When pharmacological or physiological stressors are administered, motion artefacts could be induced, as shown by Naum et al[3] in the context of physiological stress conditions. Our finding that MBF was sensitive to IF motion rather than IS motion was considered to be valid for a stress study; IF motion might be source of severe error in MBF and CFR measurements. The rigid body model, which was used in this study, has been promoted for motion compensation not only in brain PET but also in cardiac PET. McCord et al[1] employed the rigid body model to correct for misalignment during the transmission and ^{18}F-fluorodeoxyglucose (^{18}F-FDG) emission scans. Bacharach et al[2]

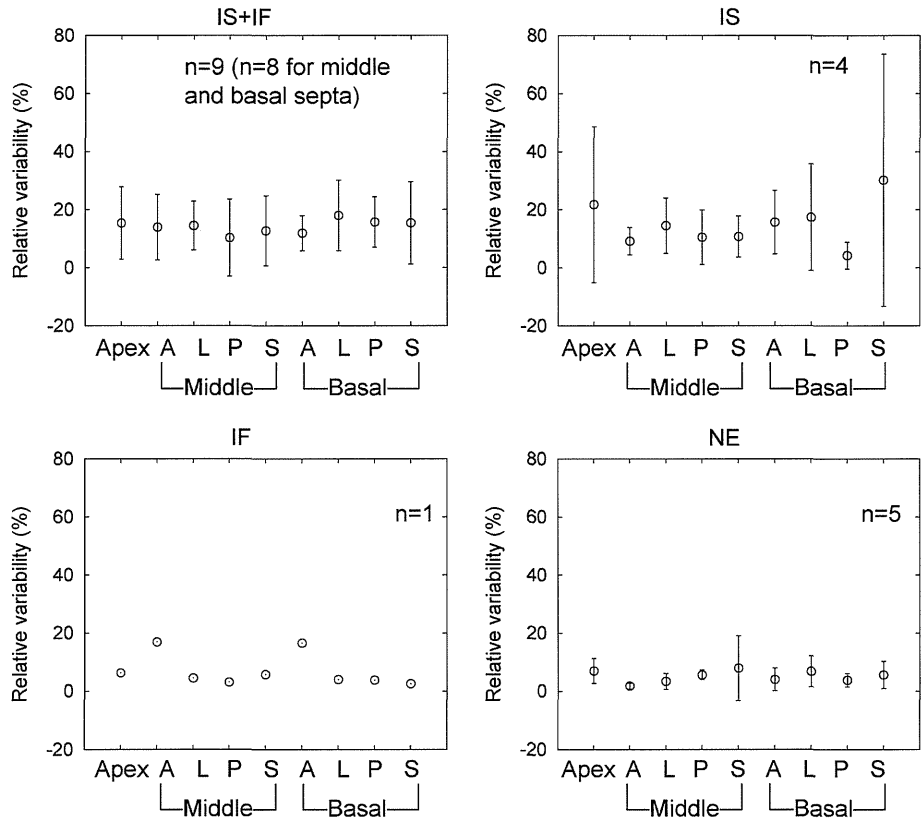

Figure 6. Relative variability of regional PTF due to effects of global motion. *IS*, Inter-scan motion; *IF*, inter-frame motion; NE represents no significant motion. IS + IF − IS and IS + IF − IF are groups corrected for IS and IF motion in the IS + IF group, respectively.

also proposed a registration technique based on the rigid body model for ^{18}F-FDG emission images acquired on different days. The use of the rigid body model is considered to be valid for non-gated cardiac PET images because the images were smoothed spatially due to cardiac wall motion, and also averaged temporally over the duration of the dynamic frames. However, in stress studies, over-correction due to relatively large respiratory motions might be introduced by our system because the subject's motion was estimated by measurement of the location of a target attached on the thoracic surface. With this in mind, further work is necessary to evaluate the contribution of MC in combination with a respiratory gating technique.

The MC system was applied to clinical follow-up studies of the patients who received LVAS and cell transplantation therapy. The aim of this study was to investigate the effects of global movement on quantification of MBF and PTF values in a single ^{15}O-water PET study under resting conditions. Although, we believed that the MC system could contribute to an accurate evaluation of regional perfusion and viability, the effects of LVAS implantation and AMS transplantation therapy on those patients is beyond the scope of the present study, and will be studied elsewhere.

CONCLUSIONS

This study demonstrated that IF motion during ^{15}O-water scans under resting conditions could be the source of error in MBF measurement. Furthermore, estimated MBF values were less sensitive than PTF values to misalignment between the transmission and the ^{15}O-water emission scans.

Acknowledgments

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Conflicts of interest

The authors have no conflicts of interest to disclose.

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Quantification of regional cerebral blood flow in rats using an arteriovenous shunt and micro-PET

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Abstract

Introduction: Measurement of regional cerebral blood flow (rCBF) in rodents can provide knowledge of pathophysiology of the cerebral circulation, but generally requires blood sampling for analysis during positron emission tomography (PET). We therefore tested the feasibility of using an arteriovenous (AV) shunt in rats for less invasive blood analysis.
Methods: Six anesthetized rats received $[^{15}O]H_2O$ and $[^{15}O]CO$ PET scans with their femoral artery and vein connected by an AV shunt, the activity within which was measured with a germanium ortho-oxysilicate scintillation detector. The $[^{15}O]H_2O$ was intravenously injected either at a faster or slower injection rate, while animals were placed either with their head or heart centered in the gantry. The time–activity curve (TAC) from the AV shunt was compared with that from the cardiac ventricle in PET image. The rCBF values were calculated by a nonlinear least-square method using the dispersion-corrected AV-shunt TAC as an input.
Results: The AV-shunt TAC had higher signal-to-noise ratio, but also had delay and dispersion compared with the image-derived TAC. The delay time between the AV-shunt TAC and image-based TAC ranged from 11 to 21 s, while the dispersion was estimated to be ~5 s as a time constant of the dispersion model of exponential function, and both were properly corrected. In a steady-state condition of $[^{15}O]CO$ PET, the blood activity concentration by AV-shunt TAC was also comparable in height with the image-based TAC corrected for partial volume. Whole-brain CBF values measured by $[^{15}O]H_2O$ were 0.37±0.04 (mean±S.D.) ml/g/min, partition coefficient was 0.73±0.04 ml/g, and the CBF varied in a linear relationship with partial pressure of carbon dioxide during each scan.
Conclusions: The AV-shunt technique allows less invasive, quantitative and reproducible measurement of rCBF in $[^{15}O]H_2O$ PET studies in rats than direct blood sampling and radioassay.

Keywords: Positron emission tomography; Oxygen radioisotopes; Rats; Cerebral blood flow; AV shunt; CO_2

1. Introduction

Regional cerebral blood flow (rCBF) is one of the most frequently used clinically evaluable parameters for characterizing regional hemodynamic and/or functional status of the brain, and can be noninvasively measured by positron emission tomography (PET) using ^{15}O-labeled water [1]. The short physical half-life of ^{15}O (~2 min) also allows repeat measurements of rCBF under different conditions within a relatively short period of time. One of the concerns when measuring rodent rCBF using $[^{15}O]H_2O$ PET is blood sampling from an artery, which is needed to obtain the arterial input function and calculate rCBF. Owing to the small total volume of blood in rodents, withdrawal of blood

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may alter the physiological conditions of the animals, resulting in instability or bias in the measured rCBF. An image-driven input function in rat PET, which derives time–activity curve (TAC) from a region of interest in the cardiac ventricle, may be a potential solution [2]. However, such image-based data may be compromised by statistical (i.e., count-related) 'noise' or biased by the partial volume effect; thus, it is not so simple to estimate arterial input function. Yee et al. recently proposed a model to estimate the input function in rats from the cardiac PET [^{15}O]H_2O image [2], but the model assumed multiple heuristic parameters and is thus difficult to validate.

The arteriovenous (AV)-shunt technique is one of the possible approaches to obtain the input function, and several studies have used this technique using radioactive tracers [3–6]. This technique involves placing a shunt between an artery and vein using a catheter and tube, in which blood activity concentration could be measured by an external detector. The advantages of the AV-shunt technique are high temporal resolution and no loss of blood during the PET study. However, there is concern of potential bias in the arterial input function, for example, due to time delay and dispersion, which may be augmented by a lengthy shunt. In addition, potential unstable flow resulting from blood clotting within the tube has not been well studied.

The aim of this study was to validate the arterial input function obtained by the AV shunt as a reference of image-based one and to quantify the rCBF of rats using [^{15}O]H_2O PET. For this purpose, we first evaluated the stability of blood flow velocity in the AV-shunt line by ultrasonography during the PET scans. Then, we separately performed two scans, one for the heart and the other for the brain, in each animal. The former scan was obtained to validate the shape of arterial input function from the AV shunt, and the latter for calculation of rCBF using the AV-shunt-based TAC. To validate the shape of the AV-shunt-derived blood TAC, the two different methods of injections (faster or slower) were performed. Thereafter, the delay and dispersion of the AV-shunt-based input function were evaluated by comparing with the input function from a region of interest in the left ventricle in the PET images. For the calculation of rCBF, the delay- and dispersion-corrected AV-shunt-derived TAC was used, and we investigated the relationship between the measured rCBF and arterial carbon dioxide partial pressure (pCO_2) during each measurement to confirm whether the current system reproduces the known physiological linear relationship between rCBF and pCO_2.

2. Material and methods

2.1. Overview of the study

Fig. 1 presents a schematic diagram of our experimental setup. To confirm the stability of the AV shunt during the PET scan, we continuously monitored the blood flow velocity in the AV-shunt line using ultrasonography. For injection of [^{15}O]H_2O, we tested two injection procedures, namely, faster and slower bolus injection, as shown in Fig. 1C. Fig. 2 shows our experimental protocol for PET scans. A total of three emission scans of [^{15}O]CO for heart, [^{15}O]H_2O for heart and [^{15}O]H_2O for brain were performed in each animal, with additional transmission scans for attenuation correction of 511 keV gamma rays. The cardiac [^{15}O]CO image was used to identify the cardiac ventricle to obtain image-based arterial TACs, while the cardiac [^{15}O]H_2O images were utilized to compare the shape of the arterial input function from the heart and from the AV-shunt line as well as to estimate the dispersion parameter of the input function from the AV shunt. [^{15}O]CO is known to label the red blood cells in the blood, and the specific activity of the supplied [^{15}O]CO was high enough to prevent side effects.

2.2. Animal preparation

Six male Sprague–Dawley rats (6 weeks old) were used in this study. Their body weight ranged from 257 to 319 g (mean±standard deviation, 283±22). The animals were maintained and handled in accordance with the *Guidelines for Proper Conduct of Animal Experiments* (June 1, 2006 Science Council of Japan). The study protocol was approved by the Laboratory Animal Welfare of the National Cardiovascular Center.

We performed all the experiments under deep anesthesia. The anesthesia was induced by intramuscular injection of ketamine (30–50 mg/kg) and xylazine (5–10 mg/kg). A rat was intubated for inhalation of [^{15}O]CO gas and for management of respiration using a respirator (SN-480-7, Shimano, Tokyo, Japan). Two 22-gauge needles were percutaneously inserted into the caudal veins and were attached to 25-cm-long polyethylene tubes for maintenance anesthesia using propofol (20–25 mg/kg/h) and for injection of [^{15}O]H_2O. The right femoral artery was catheterized with a polyethylene tube (PE50, inner diameter of 0.54 mm) for sampling blood for blood gas analysis and for monitoring heart rate and blood pressure using a physiological monitoring device (BSM-2391 Nihon Kohden, Tokyo, Japan). The left femoral artery and vein were also catheterized with PE50 tubes for the AV shunt. The length of the tubes was 55 and 35 cm for the left femoral artery and left femoral vein, respectively, and 10 cm of the ends of the two lines was connected to a thicker polyethylene tube (PE 100, inner diameter of 0.86 mm) to form the AV shunt and to allow passive blood flow dependent on the pressure gradient from artery to vein. The thicker part of the tube was used for monitoring blood flow by the ultrasonographic probe as described below, and a part of the arterial side of the PE50 tube (35–45 cm from the femoral artery) was used for online measurement of activity concentration using a Gd_2SiO_5:Ce (GSO) detector (GSO input function monitor system, Molecular Imaging Laboratory, Osaka, Japan) [7]. The GSO detector consists of two pairs of GSO crystals (20 mm×20 mm×12.9 mm for each) with photomultiplier tube

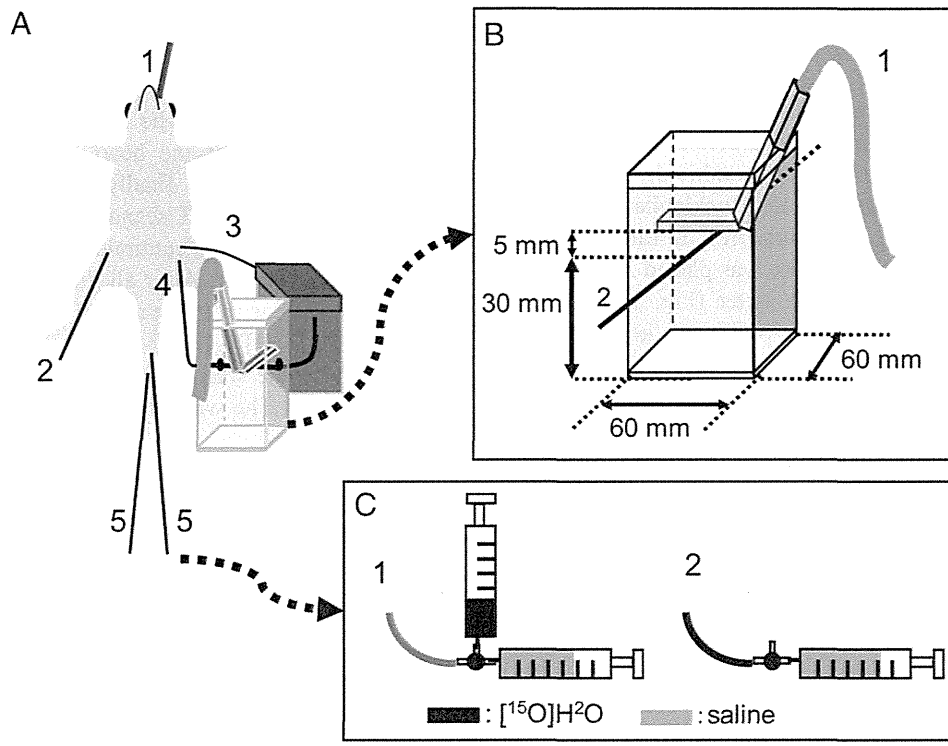

Fig. 1. A schematic diagram of the experimental setup. (A) Animal preparation: intubation for inhalation (1), blood catheter lines in the left femoral artery (3) and vein (4) for AV shunt, and in the right femoral artery (2) for monitoring heart pressure, and in two caudal veins (5), one for injection of $[^{15}O]H_2O$ and the other for infusion of anesthetics. (B) Measurements of blood flow velocity in the AV-shunt line by ultrasonography: the appliance consisted of an echo probe (1) and AV-shunt line (2), and the container was filled with water. The distance between the AV-shunt line and the echo probe was kept at 5 mm. (C) Two injection methods: faster bolus injection (1) and slower bolus injection (2). In the faster bolus injection, a solution of $[^{15}O]H_2O$ (volume of ~0.3 ml) in a 1-ml syringe was injected in a period of 15 s into the tail vein, and then the three-way stopcock was turned to the other syringe to flush through 0.7 ml of saline. In the slower bolus injection, $[^{15}O]H_2O$ (~0.3 ml) filled the venous catheter line before injection and was flushed through by 0.7±0.1 ml of saline in the syringe over a period of 30 s.

and coincidence circuits. A 2-cm-thick lead is utilized for shielding. The absolute sensitivity for PET tracers is approximately 7%.The total volume in the AV-shunt line was 0.26 ml (as a reference, the total amount of blood per body is ~20 ml). The AV-shunt line was coated with heparin to prevent clot formation. Soon after equipment setup, the animal was fixed in a supine position on the micro-PET imaging bed. A blood sample from the femoral artery was analyzed for pCO_2 using a blood gas analyzer (ABL-500, RADIOMETER, Tokyo, Japan). Body temperature was measured with a probe placed in the rectum, and kept at 36°C±1°C by using an infrared thermostat.

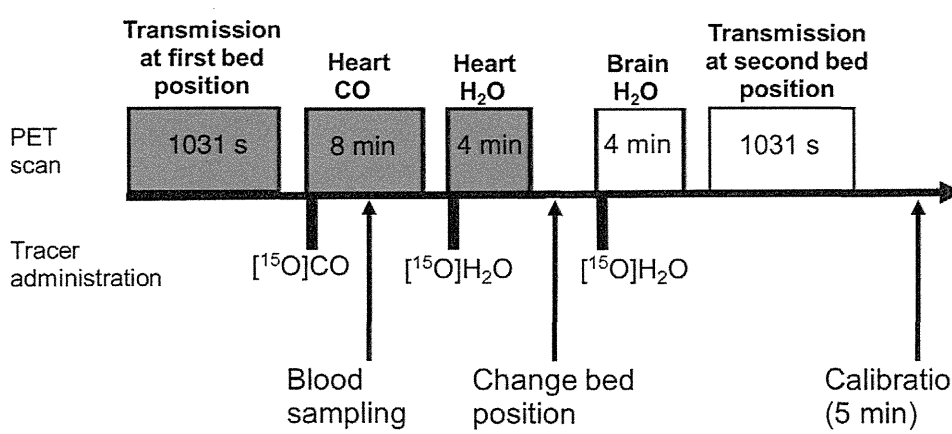

Fig. 2. Experimental protocol. The first three PET scans (gray) were performed for derivation of the cardiac ventricular TAC, while the latter two (white) were performed for brain imaging and the cerebral TAC. The activity concentration of the AV shunt was monitored and recorded to obtain the TAC during the PET scan.

2.3. Measurement of AV-shunt line blood flow velocity by ultrasonography

To confirm that blood constantly flows in the AV-shunt line, we performed measurements of blood flow velocity in three rats using a portable ultrasonographic device (LOGIQ BOOK XP, GE Healthcare UK Ltd., Buckinghamshire, UK) with a probe (i12L-RS, GE Healthcare UK Ltd., Buckinghamshire, UK). The AV-shunt line was placed 30 mm from the bottom of a box filled with water (Fig. 1B), and the distance between the line and the probe was kept at 5 mm. Blood flow velocity (cm/s) was monitored by the ultrasonographic probe during the experiment. The values of blood flow velocity measured by pulse Doppler were recorded twice during each emission scan. The velocity was scaled by the initial value measured before the PET scan of $[^{15}O]CO$ in the heart.

2.4. Injection methods

Intravenous injection of $[^{15}O]H_2O$ was performed in two ways, according to injection rate (faster, n=3 or slower, n=3) (Fig. 1C). In the faster bolus injection, a solution of $[^{15}O]H_2O$ (59±1.6 MBq, in a volume of ~0.3 ml) in a 1-ml syringe was injected in a period of 15 s into the tail vein, and then the three-way stopcock was turned to the other syringe to flush through 0.7 ml of saline. In the slower bolus injection, $[^{15}O]H_2O$ (~0.3 ml) filled the venous catheter line before injection and was flushed through by 0.7±0.1 ml of saline in the syringe over a period of 30 s. In both methods, the total injected radioactivity at the time of tracer injection was 68.6±3.3 MBq.

2.5. PET experiments

We used a PET scanner (micro-PET Focus 120, SIEMENS Healthcare USA Inc., Malvern, PA, USA) for imaging activity concentration distribution. This PET scanner is dedicated for small-animal use and has a transaxial field of view (FOV) of 10 cm and an axial FOV of 7.6 cm. The sensitivity and the spatial resolution at the center of the scanner are 7% and 1.18 mm full-width half-maximum, respectively [8]. All PET emission data were acquired in list mode.

For the scan protocol (Fig. 2), the animal was first positioned with its chest (heart) centered in the FOV of the PET scanner to scan the heart. After a transmission scan for a period of 1031 s using a $^{68}Ga/Ge$ pinpoint source, an emission scan was initiated immediately after completion of inhalation of $[^{15}O]CO$ (350±20 MBq/min) for 10 s and continued for a period of 8 min. A blood sample was taken from the femoral artery at 4–5 min after starting inhalation of $[^{15}O]CO$ and was radioassayed using a well counter (BeWell, Molecular Imaging Laboratory, Osaka, Japan). After radioactive decay, an emission scan for a period of 4 min was performed immediately after injection of $[^{15}O]H_2O$ by either faster or slower bolus injection. Then, the animal was repositioned with its head centered in the FOV of the PET scanner for brain scanning, and a 4-min emission scan was performed after injection of $[^{15}O]H_2O$ by either faster or slower bolus injection. The injection method of $[^{15}O]H_2O$ was the same as for the heart scan for each subject. A transmission scan was finally acquired for correction of attenuation. During PET scanning, activity concentration in the AV-shunt line was continuously recorded every second by the GSO detector. After all the PET scans, we removed the AV-shunt line from the animal, filled the line with saline containing $[^{15}O]H_2O$ of known activity concentration measured by the well counter, for cross-calibration of the GSO detector. The well counter was also calibrated to the PET scanner.

PET data, acquired in list mode, were sorted into multiple frames of 3D sinogram data. The emission data for $[^{15}O]CO$ consisted of 31 frames (18 frames×10 s, 12 frames×20 s, 1 frame×60 s), and for $[^{15}O]H_2O$ (in both heart and brain scans), 63 frames (30 frames×2 s, 30 frames×5 s, 3 frames×10 s). All sinogram data were reconstructed by a filtered back projection algorithm (a Ramp filter, cutoff frequency of 0.5 of the Nyquist frequency) with Fourier rebinning and with correction of attenuation using transmission data.

The TACs of arterial blood in the AV shunt were obtained from activity concentration data measured by the GSO detector, corrected for physical decay of ^{15}O and for cross-calibration between the GSO detector and the well counter.

2.6. Determination of image-derived TACs

To derive the image-based TAC, we identified the left ventricular blood pool from multiple PET images and derived time–activity concentration data from the H_2O PET scan [9]. To identify the region of blood pool, we subtracted the summed $[^{15}O]CO$ image from the summed $[^{15}O]H_2O$ image to generate an image that more clearly delineated the border of the myocardium and ventricle (Fig. 3A). A volume of interest (VOI) was defined by manually drawing the border of the ventricle, and the VOI was superimposed on the $[^{15}O]H_2O$ dynamic images to obtain the TAC. As shown in this figure, the initial phase of the $[^{15}O]H_2O$ image showed higher accumulation in the ventricle, while at the later equilibrium phase, the ventricular blood pool became indistinguishable from the myocardium, suggesting more homogenous activity concentration between these compartments at this phase. For the VOI analysis, we used the ASIPro VMimage analysis software (Concorde Micro systems, Knoxville, TN, USA).

2.7. Consistency of input functions obtained from the AV shunt and PET image

To compare the shape of the TACs from the AV-shunt technique and the PET image, we first derived an averaged TAC for each injection method (faster vs. slower) and measurement modality (AV shunt vs. image-derived). Averaging was performed across measurements and animals

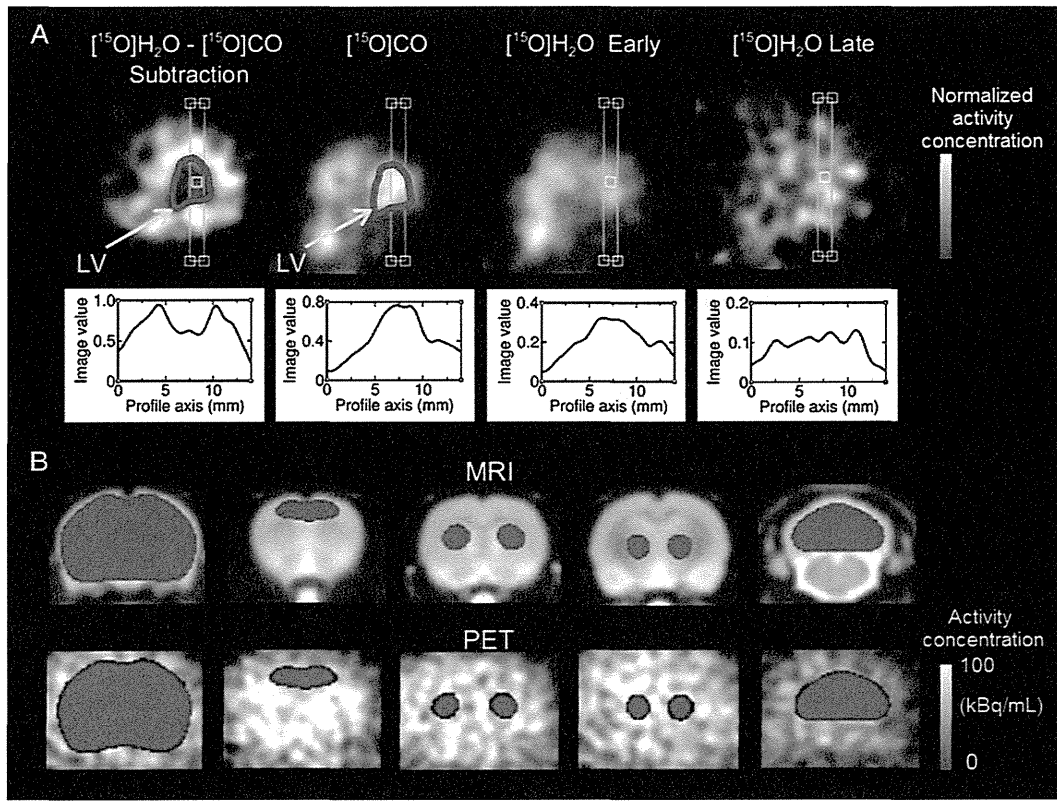

Fig. 3. VOIs. (A) VOI in the cardiac ventricle (delineated by red line). The VOI, delineated manually using the subtraction image of [^{15}O]H_2O minus [^{15}O]CO activity concentration (upper left), is transferred to the sum image of [^{15}O]CO [^{15}O]H_2O in the early phase (0–30 s) and [^{15}O]H_2O in the late phase (150–180 s). Lower graphs show the profile of activity concentration on the line that is crossing the center of cardiac ventricle (as shown in light blue overlaid on the upper PET image). The image value in the graphs is in units of MBq/ml. Each PET image is shown in a gray color bar in an arbitrary range. (B) The coronal sections of MRI (upper) and PET images (lower) on which cerebral VOIs are overlaid in red. The furthest left section shows the whole-brain VOI. From the second left to right, the distance of each coronal slices was 2, 0, −2.6 and −12.0 mm from the bregma, each depicting the VOIs in the frontal cortex, striatum, thalamus and cerebellum, respectively. Each PET image was obtained by averaging dynamic PET images from 0 to 3 min. The activity concentration shown ranged from 0 to 100 kBq/ml as indicated in the gray color bar.

after normalization of the height and area under the curve. To eliminate variations in administered activity in each measurement, TACs were divided by the area under the curve (from 0 to 180 s). The rising time of each TAC was shifted to time zero by visual inspection. Then, the mean and standard deviation of the TACs were separately determined for each injection method and measurement modality. Finally, averaged TACs were normalized by their peak height and presented in the line chart (in Fig. 4A–B). A pair of AV-shunt and image-based TACs was obtained from the same PET measurement.

Since visual inspection of the shape of the average TAC led us to suspect significant dispersion in AV-shunt TAC as compared with image-based TAC, we investigated whether the dispersion could be corrected before calculating quantitative rCBF [9]. The correction of dispersion can be achieved by an empirical model of dispersion in which dispersion function was expressed in monoexponential function [9]. We estimated a dispersion time constant in this model equation using the "averaged TACs" of AV shunt and image based, since a statistical error in the TAC may affect the estimation of the dispersion. Dispersion time constants (τ) of 0, 5 and 10 s were used to derive deconvolved TACs from the mean AV-shunt TACs and were plotted against time along with the image-based mean TAC. Among three tested values for τ, we chose the optimal dispersion time constant for the AV-shunt TAC by visually inspecting consistency of the shape of TACs during the initial rising phase (faster: 0–15 s, slower: 0–30 s) between deconvoluted AV-shunt and the image-based TACs. We used a program for deconvolution written in the C programming language.

Then, we compared the "absolute" activity concentration of the TACs of the AV-shunt and image-derived techniques, both obtained simultaneously in the same animal and PET scan. This was primarily for the purpose of determining the adequacy of the calibration and confirming the suitability of the AV-shunt TAC using image-based TAC as a reference. As described previously, the activity concentration of the image-based TAC obtained from cardiac ventricle should reflect the sum of the underestimated activity concentration in the cardiac ventricle and spillover activity from surrounding tissues (cardiac wall tissues). Thus, we estimated the intraventricular blood activity concentration from imaged-

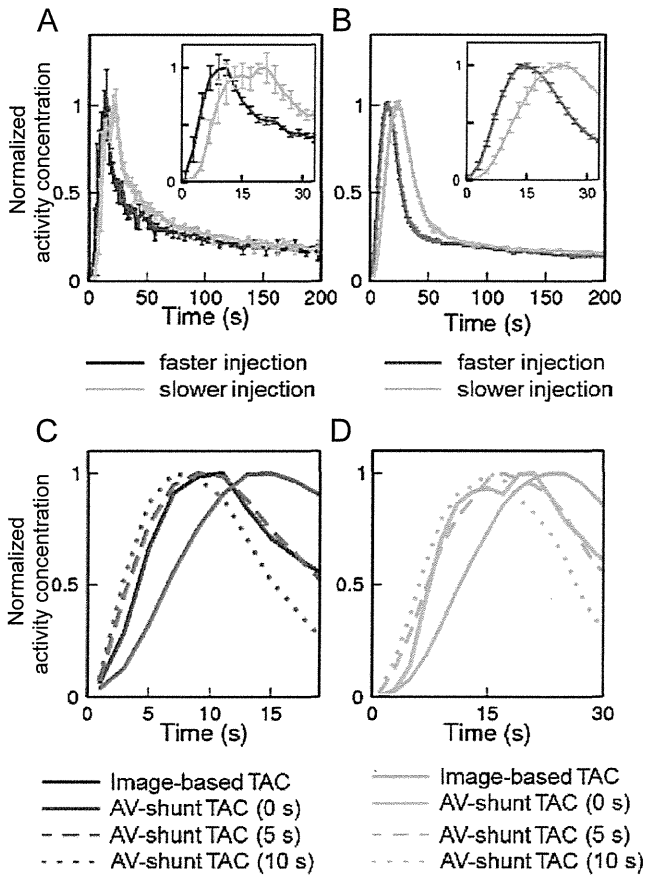

Fig. 4. Comparison of the normalized imaged-based and AV-shunt-derived TACs and of their dispersions. (A) Mean and standard deviation of normalized imaged-based TAC (obtained from the cardiac ventricular VOI) (n=3). Dark blue curve, faster injection; light blue curve, slower injection. (B) Mean and standard deviation of normalized AV shunt TAC (n=3). Red line, faster injection; orange line, slower injection. In each figure, the error bar indicates standard deviation, and the right upper inset shows the initial rising phase for a period of 30 s. Normalization was performed by scaling using the area under the curve and then the peak of TAC. Note that the standard deviation, particularly in the early rising phase, was much smaller in the AV-shunt TAC as compared with the image-based TAC. (C) Dispersion correction of AV-shunt TAC in faster injection. The blue line indicates the mean TAC obtained from the cardiac ventricle in the PET image, while red lines indicate the TAC of the AV shunt after correction of dispersion with time constants of 0, 5 and 10 s, respectively. (D) Dispersion correction of AV-shunt TAC in slower injection. The light blue line indicates the mean TAC obtained from the cardiac ventricle in the PET image, while orange lines indicate the TAC of the AV after correction of dispersion with time constants of 0, 5 and 10 s. In both faster and slower injection, the dispersion time constant of ~5 s appears to be the most optimal among the three for correcting the dispersion of AV-shunt TAC.

based TAC using a one-tissue, one-blood compartmental model of partial volume for the cardiac VOI [10]. The activity concentration in the cardiac ventricular VOI, $C_v(t)$ (Bq/ml), can be expressed by the following equation:

$$C_v(t) = \beta C_a(t) + \gamma\rho C_m(t), \quad (1)$$

where β is a recovery coefficient, γ is the spillover fraction of tissue activity concentration into left ventricle VOI, ρ is the density of myocardium (g/ml) and C_m the activity concentration of myocardium (Bq/g) [10]. Here, $\beta+\gamma=1$ if the heart (i.e., the dimensions of the left-ventricular cavity and myocardium) is "large" compared to the spatial resolution of the PET system [10]. The recovery coefficient (β) can be measured by calculating a ratio between mean activity concentration of the cardiac ventricular VOI during steady-state condition in the [^{15}O]CO scan (Fig. 3) and actual blood activity concentration sampled during that period, as described previously [11,12]. In addition, under an equilibrium state, the activity concentration of myocardium (C_m) can be estimated by the following equation:

$$C_m = pC_a, \quad (2)$$

where p is a partition coefficient of cardiac tissue (ml/g) and C_a is the activity concentration of [^{15}O]H_2O in blood (Bq/ml); thus, using Eqs. (1) and (2), C_a at an equilibrium state can be estimated from measured C_v by the following equation:

$$C_a = C_v/[\beta + (1-\beta)\rho p] \quad (3)$$

using known values of ρ and p, 1.04 g/ml and 0.91 ml/g, respectively [10]. Therefore, the comparison of the C_a values estimated from Eq. (3) with the activity concentration of AV shunt at steady state (this should correspond to the blood activity concentration) will mainly indicate the eligibility of the two techniques. Since the equilibrium was achieved at time t from 120 to 180 s after PET scan, the value of C_v was measured from the cardiac ventricular VOI by averaging the activity concentration of [^{15}O]H_2O during the corresponding period, and C_a was estimated from Eq. (3) using the measured value of β. The activity concentration of the AV shunt was obtained from averaged activity concentration during a period of 120–180 s and was compared with the values of C_a predicted from Eq. (3) and the image-based (cardiac ventricular VOI) TAC.

2.8. Regional CBF determination

Cerebral tissue TACs during [^{15}O]H_2O PET scans were obtained from five regions of interest in the brain: frontal cortex, striatum, thalamus, cerebellum and whole brain. A standard T2-weighted magnetic resonance image (MRI) of a rat [13] was automatically realigned to the summed PET image by means of an image registration tool in the FSL software (version 4.1.4) (http://www.fmrib.ox.ac.uk/fsl/index.html) [14]. The cross-modal image registration was performed using a cost function of mutual information. The VOI for each brain region was manually drawn on the standard MRI, realigned using an inverted transformation matrix to the PET image and used to obtain the TAC for each VOI (Fig. 3B). The VOI volumes were 1472, 44.0, 22.4, 13.5 and 106 mm^3 for the whole brain, frontal cortex, striatum, thalamus and cerebellum, respectively. The rCBF values were determined by using the input function of the

AV-shunt and cerebral TAC data in two steps. In the first step, we determined the time delay of the AV-shunt TAC as compared with cerebral whole-brain TAC by visual inspection. Using the predetermined time delay, the rCBF value for each region was estimated by means of a nonlinear least-square fitting procedure based on a one-tissue compartmental model [1–9,11–15], with unknown parameters of rCBF and partition coefficient p [16–18]. For the cerebral TAC, the initial period of 200 s was used for the fitting. The input function from the AV shunt was corrected for dispersion by deconvolving the original TAC using a single exponential function with a time constant of 5 s, determined based on the previous analysis.

Mathematical manipulations of TAC data and the fitting process to determine CBF were all performed using PyBLD (http://www.mi.med.osaka-u.ac.jp/pybld/pybld.html) built upon the Python language.

3. Results

3.1. Measurements of blood flow velocity in the AV-shunt line by ultrasonography

We evaluated the stability of the blood flow velocity in the AV-shunt line by comparing the relative velocity with that measured at a reference time point just before the [^{15}O] CO scan (Fig. 2). The time interval between the reference and the last time point (fourth) was 62±9 min. When scaled by the velocity at the reference point, the relative velocity at the later time ranged from 0.9 to 1.3 (Table 1), and there was no single case in which the flow velocity in the AV-shunt line was systematically biased during the experiments.

3.2. Comparisons of shape of TACs between different methods of injection and input measurement

Fig. 4A–B shows the average and the standard deviation of TACs in which the peak and area under the curve were normalized and averaged for each injection method (faster vs. slower) and measurement method (PET image based vs. AV shunt and GSO based). As predicted, the faster bolus injection showed an earlier rising time and a sharper initial phase of the TACs than after slower injection in both measurement methods. The image-based method provided a much faster and more rapid increase in the normalized activity and showed a sharper spike shape of the TAC, suggesting less dispersion of the curve shape. However, the standard deviation of the image-based TAC was relatively larger than that of the AV-shunt TAC, suggesting that the image-derived TACs suffer from more statistical noise and/or net detector sensitivity than the AV-shunt method. As also shown in Fig. 4, the AV-shunt-derived TACs were more dispersed than the image-derived converts, particularly in the initial portion of the curves, suggesting that a correction for the dispersion of AV-shunt-derived TACs may be required. The estimated time delays for the AV-shunt-derived versus the image-based TACs are listed in Table 2.

Table 1
Relative velocity of blood in the AV shunt during PET scan

	PET scan		
	Heart [^{15}O]CO	Heart [^{15}O]H_2O	Brain [^{15}O]H_2O
Relative velocity	1.0(0.1)	1.1 (0.1)	1.2 (0.1)

Values were scaled by the initial velocity measured before heart [^{15}O]CO PET scan. Number of observation was n=6 for each variable. Values were shown as mean (S.D.). The repeated-measure ANOVA did not show any significant effect of scans.

Table 2
Estimated time delay of AV-shunt TAC as compared with image-based TAC

Injection methods	Estimated delay (s)	
	Between AV-shunt TAC and image-based TAC of cardiac ventricle	Between AV-shunt TAC and whole-brain TAC
Faster injection	11 (9, 12)	9 (7, 11)
Slower injection	20 (18, 21)	17 (16, 18)

Values are shown in median (minimum, maximum) in unit of second. The number of observation was n=3 for each injection method. Two-way ANOVA with factors of injection speed (faster vs. slower) and reference TAC (image-based TAC vs. whole-brain TAC) disclosed significant effects of injection speed ($F_{1,8}$=89.7, P<.001) and reference TAC ($F_{1,8}$=5.82, P<.05).

3.3. Dispersion correction of AV-shunt TAC

Fig. 4C shows the effect of dispersion correction on averaged TACs. When corrected by τ=5 s, the shape of the dispersion-corrected mean AV-shunt TAC agreed well with that of the image-derived TAC, as compared with those with corrections of τ=0 and 10 s. We did not estimate the degree of dispersion for each single measurement since the deconvolution performed in the dispersion correction enhances the statistical noise of TAC curves, making it difficult to estimate the size of dispersion. Therefore, we applied the fixed value of τ=5 for correcting the dispersion for each AV-shunt TAC and used them for further analysis since the dispersion constant may be considered almost invariable if the geometry of blood flow paths including arterial architecture and AV-shunt line were the same across measurements.

3.4. Absolute activity concentration in the AV-shunt and image-derived TACs

Fig. 5A shows a comparison of the absolute activity concentration of the TACs obtained from the AV shunt and cardiac PET image in all the test animals. As shown in this figure, at the early phase, the peak of the AV-shunt TAC (red line) was higher than that of image-based TAC (blue line). As shown in Fig. 3A (in the panel second from the right), at the early phase of [^{15}O]H_2O scan, the activity concentration of the [^{15}O]H_2O was higher in the center of cardiac ventricle,

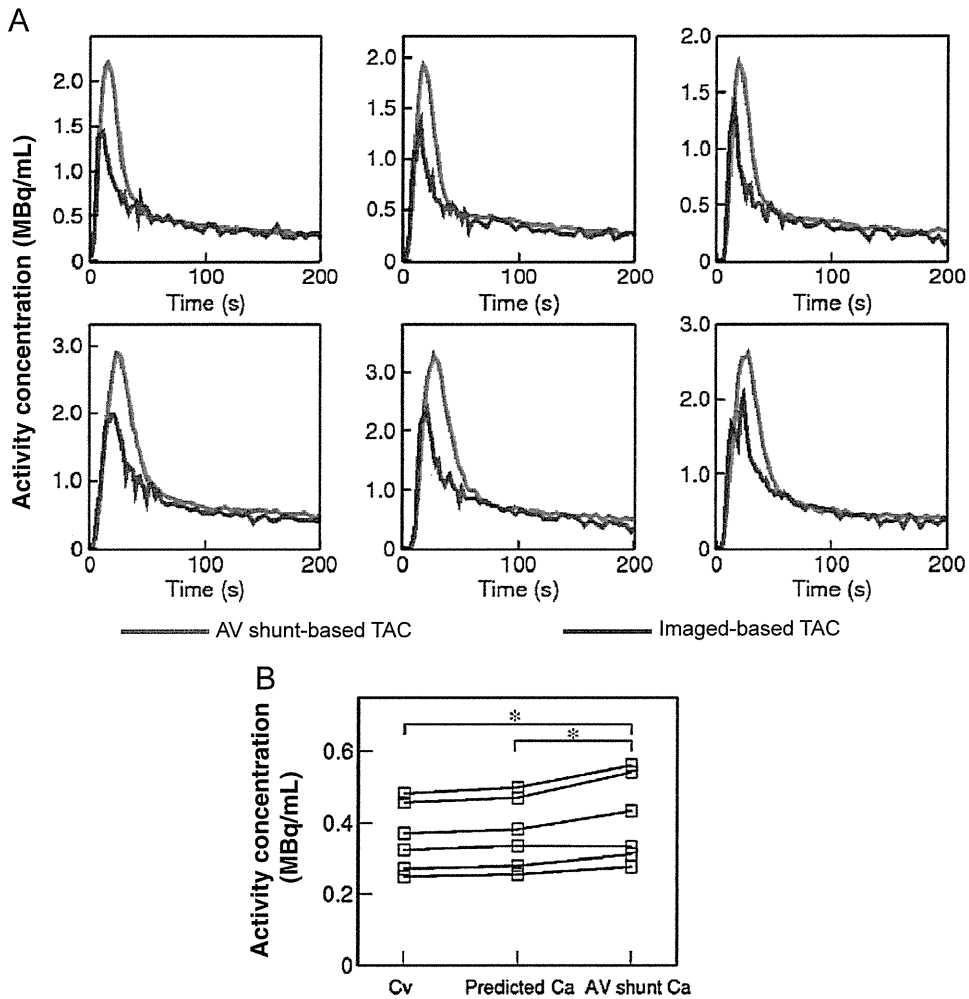

Fig. 5. Image-based and AV-shunt-derived TACs. (A) Absolute activity concentration of the AV-shunt TAC (red) and image-based cardiac ventricular TAC (blue) in all animals. The AV-shunt TAC (red) was corrected for dispersion (a time constant of 5 s), delay and calibration between the GSO detector and PET. Note that the image-based TAC shows almost comparable activity concentration as the AV shunt in the later equilibrium stage (from ~50 to 180 s), while it showed lower peak height of activity concentration in the early phase. (B) Activity concentration measured in the image-based cardiac ventricular VOI (C_v), predicted blood activity concentration (predicted C_a) and measured AV-shunt activity concentration (AV-shunt C_a). *P<.05 in paired t test.

suggesting the underestimation of blood activity concentration in the image-based TAC. Actually, the value of β (recovery coefficient for the cardiac VOI) was evaluated to be 0.43±0.05 from the data obtained during [^{15}O]CO scan (Table 3). The underestimation of the peak concentration was ~28.3%±4.4% by the image-based TAC as compared with AV-shunt TAC. In addition, at the later phase, the activity concentration of image-based TAC was comparable to the height of AV-shunt TAC in all animals (Fig. 5A),

Table 3
Measured and estimated variables from VOI of left ventricle

Variables	Mean±S.D.
VOI (μl)	29.0±0.3
Estimated recovery coefficient of left ventricular VOI (β)	0.43±0.05

Values were obtained from cardiac PET data with [^{15}O]CO and [^{15}O]H_2O (n=6).

suggesting that [^{15}O]H_2O diffused into the cardiac wall tissue and the spatial distribution of the activity extended over a larger region including the myocardial wall as well as the intraventricular cavity (i.e., a region whose dimensions were sufficiently large to minimize the partial volume effect). Consistent with this hypothesis, the count profiles over the late [^{15}O]H_2O images were flatter than those over the early images (Fig. 3A).

The foregoing was corroborated by comparison of the steady-state portions of the AV-shunt-derived TAC and of the imaged-derived curve based on the one-tissue, one-blood compartment model [Eq. (3)]. The averaged blood activity concentration (120<t<180 s) measured by the AV shunt varied from 0.28 to 0.56 MBq/ml (mean±S.D.=0.41±0.11 MBq/ml). The predicted blood activity from Eq. (3) using the image-based VOI data ranged from 0.25 to 0.50 kBq/ml (0.37±0.09 MBq/ml). Fig. 5B shows a plot of the measured image-based VOI activity concentration (C_v), blood activity

concentration predicted by Eq. (3) (predicted C_a) and AV-shunt blood activity concentration(AV-shunt C_a) in each animal. As shown in this figure, there was significant bias among three measures [repeated analysis of variance (ANOVA), $F_{2,\ 5}$=17.9, P<.001]. There was significant differences in blood activity concentration between C_v and AV-shunt C_a and between the predicted C_a and the AV-shunt C_a (post hoc pairwise comparison with correction multiple comparison, P<.05). There was a significant linear relationship between predicted C_a and AV-shunt C_a [AV-shunt C_a (MBq)=1.21∗predicted C_a (MBq)−4.0×10^{-2}, R^2=0.98, P<.001]. This suggests, however, that the bias was relatively small (i.e., only ~20%); the measured activity concentration of AV shunt at the steady state was comparable to and consistent with the blood activity concentration predicted from image-based VOI in the current system of rat heart/micro-PET.

3.5. Quantification of rCBF

The mean and S.D. of rCBF values (in ml/min/g) was 0.37±0.04 in the whole brain, 0.40±0.08 in the frontal cortex, 0.41±0.06 in the striatum, 0.36±0.04 in the thalamus and 0.36±0.05 in the cerebellum (see Table 4 for all the data). The mean and S.D. for partition coefficient (p) in the whole brain was 0.73±0.04 ml/g. No statistically significant difference was observed in rCBF values between faster and slower bolus injections (P>.05). Whole-brain rCBF values were plotted against pCO_2 values in all the animals except one, in which we failed to measure the pCO_2 (Fig. 6). As shown in this figure, a linear relationship was obtained between pCO_2 and whole-brain rCBF values [rCBF=0.01×pCO_2+0.06 (R^2=0.90, P<.001)]. The estimated time delay between whole-brain TAC and AV-shunt TAC is also listed in Table 2; as predicted, the estimated delay was consistently 3–4 s shorter than that between cardiac ventricular TAC and AV-shunt TAC for both injection methods.

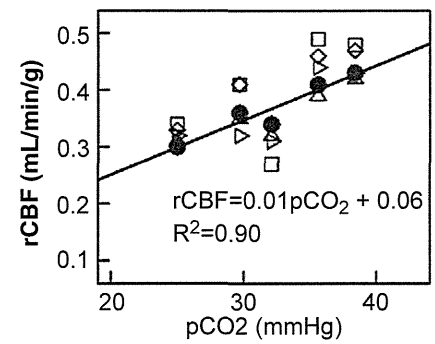

Fig. 6. Correlation between rCBF by brain region and blood pCO_2 using the AV-shunt-derived arterial input function. ●: whole brain, □: frontal cortex, ◇: striatum, Δ: thalamus, ▽: cerebellum.

4. Discussion

There has been no unbiased measurement method for arterial input function (as well as for blood flow) in rodents PET. The manual sampling may have been a conventional way of measuring input function, but apparently, it should have substantial bias due to insufficient temporal resolution. Invasiveness of this method makes it difficult for us to fully analyze or understand the actual bias of this system. In this paper, we described an AV-shunt technique for acquiring the arterial input function of [^{15}O]H_2O during PET acquisition in rats and used an image-based method as a reference of AV-shunt since it has already been used in the past literatures [2,10]. In the ultrasonographic evaluation of the blood flow through the AV shunt, none of the animals showed any evidence of stagnation in the shunt line, which potentially degrades the measurement of blood activity or the shape of AV-shunt TAC. The image-based TAC was obtained using the [^{15}O]H_2OPET data of the heart chamber based on a method described previously [10]. The results revealed that the count noise, one of the major error sources of activity concentration, was small in the AV-shunt TAC (as compared with image-based TAC) and that the shape of the AV-shunt-based TAC is not attenuated very much and could be properly corrected for time delay and dispersion. In contrast, the image-based method suffered from the partial volume effect even after the model-based correction, and the effect size was varied across time after injection. The PET scan for the brain with the AV shunt provided rCBF values that showed physiologically plausible variation depending on the blood carbon dioxide level. These findings suggest that the AV-shunt technique could be useful for future studies for rCBF measurements. The concept of the current study in

Table 4
rCBF values measured by [^{15}O]-H_2O PET and AV-shunt technique

Subjects	pCO_2	rCBF (p)				
		Whole brain	Frontal cortex	Striatum	Thalamus	Cerebellum
1	32.1	0.34 (0.77)	0.27 (0.64)	0.34 (0.86)	0.32 (0.94)	0.31 (0.76)
2	38.4	0.43 (0.80)	0.48 (0.88)	0.47 (0.89)	0.42 (0.86)	0.43 (0.78)
3	35.6	0.41 (0.71)	0.49 (0.80)	0.46 (0.79)	0.39 (0.96)	0.44 (0.72)
4	–	0.37 (0.70)	0.44 (0.77)	0.40 (0.79)	0.37 (0.77)	0.36 (0.66)
5	25.0	0.30 (0.69)	0.34 (0.71)	0.33 (0.73)	0.31 (0.74)	0.32 (0.72)
6	29.7	0.36 (0.69)	0.41 (0.73)	0.41 (0.78)	0.35 (0.80)	0.32 (0.65)

The values of pCO_2 are shown in unit of mmHg, rCBF in ml/g/min and p in ml/g, respectively. We failed to measure the value of pCO_2 of subject 4.

terms of bias and noise in measured arterial inputs is not new in the field of clinical nuclear medicine, but has been rarely taken into account in the field of molecular imaging in small animals. The results provide important messages for researchers who quantify CBF in rodents using PET.

One advantage of the AV-shunt TAC method over the image-based one may be higher temporal resolution and signal-to-noise ratio. As shown in Fig. 4, our approach for the AV-shunt-based TAC clearly detected different shapes of the input function between the faster and slower bolus injections, despite a small difference in the total period for injection. In addition, the figure shows that the error of the mean is also relatively smaller with the AV-shunt technique compared with the image-based one. These findings indicate that the AV-shunt-derived TAC may have less statistical error, mainly due to the high sensitivity of the detector geometry, and thus yield less propagated error associated with the calculated rCBF values than the image-based TAC. The shape of AV-shunt TAC may have been degraded by the dispersion; however, the amount of the dispersion was not too large ($\tau = 5$s), which was consistent for both injection methods (faster and slower) and was accurately correctable [19]. Iida et al. showed that, in humans, the arterial TAC derived from the radial artery was comparable to the cardiac image-derived TAC if the arterial TAC was corrected for dispersion using a time constant of 4 s [20]. In rats, Weber et al. estimated that the potential dispersion of TAC originating within the shunt tube was in a very small range, but they did not evaluate the actual dispersion time constant needed for correction. Thus, it is notable that our data show that the time constant of dispersion is small enough to properly correct dispersion of AV-shunt TAC in the current rat PET system.

Another notable advantage of the use of AV-shunt-based TAC over the image-based TAC is no necessity for correction of the height of the curve other than calibration for the activity concentration detectors. The activity concentration of cardiac ventricular VOI in the early phase of PET scan consistently showed lower values than those in the AV-shunt TAC, presumably due to predominant spill-out effect rather than spill-in in this early phase. Actually, the current data of [^{15}O]CO scan indicate that this underestimation was ~0.43 as a recovery coefficient (Table 3). The steady-state blood activity concentration estimated by AV shunt (AV-shunt C_a) was very close to that predicted from the image-based TAC.

Although we used imaged-based TAC as a reference, after correction for partial volume, it did not fully recover the blood activity concentration of [^{15}O]H_2O, particularly at the early phase of TAC. There was a significant underestimation in the initial phase (~28%) and even in the steady-state period (~20%) (Fig. 5A). The reason for this underestimation may be due to the small size of the heart as compared with the resolution of the current PET scanner, which violates the assumption of the model equation for partial volume (i.e., $\beta+\gamma=1$). Actually, our [^{15}O]CO results showed small value of recovery coefficient (~0.43) for the left-ventricle ROI, and previous simulation studies also indicated that recovery coefficient of less than 0.65 may result in underestimation of TAC when derived from human left ventricle [10]. Thus, future studies should improve the model to obtain accurate blood TAC from PET image in rat heart.

Other investigators have attempted to derive to a standard input function for small-animal PET study but for a more slowly cleared and localizing tracer, namely, ^{18}F-fluorodeoxyglucose [21]. Because of the fast kinetics of [^{15}O]H_2O, it may be difficult to use the standard input function for this tracer; thus, it requires to be evaluated in detail. In particular, the difference in the delay time of the rising time of TAC should be strictly standardized. In addition, the injection rate of tracer activity concentration should always be standardized, so that an automatic injector that allows a constant infusion of the tracer is recommended. Further, our results suggest that, with careful standardized injections, the use of average small-animal arterial input functions for "fast" tracers such as [^{15}O]H_2O as well as "slower" tracers such as [123I]iodoamphetamine may be valid.

Our motivation to measure blood flow velocity in the AV shunt by ultrasonography was concern for potential occlusion of the line by blood clotting during experiments, which may affect the shape of arterial input curves. Weber et al. showed that there were rare cases of clogging in their AV-shunt method where they did not perform heparinization beforehand and used multiple stopcocks for blood sampling, which may have induced turbulent blood flow and facilitated clot formation [5]. In our case, the AV-shunt line and the animals' blood were both heparinized, and no three-way stopcock was used in the AV-shunt line, thereby avoiding any potential clot inducing of blood flow. The flow velocity inside the tube did not actually decrease within a period of about 1 h during the experiments, suggesting that clot formation was successfully prevented by the current method. On the other hand, there may be potential bleeding when heparin is administered, but none of the test animals in the current study showed such bleeding during the experiments.

There is an extensive literature on the relationship between rCBF and pCO_2. Our study showed a significant correlation between rCBF and pCO_2; in particular, the slope of the regression line ($\Delta CBF/\Delta pCO_2$=0.01) was comparable to that obtained in primate animals by Grubb et al. [22]. By an intracarotid injection method using [^{15}O]H_2O tracer and [^{15}O]CO, they showed that the rCBF values of anesthetized monkeys had the following relationship with pCO_2: rCBF (ml/g/min)=0.018×pCO_2 (mmHg)−0.17. Other researchers also reported that, in anesthetized monkeys, the value of $\Delta CBF/\Delta pCO_2$ was 0.011 [23] and 0.012 [24]. In recent literature, Williams et al. (1992) reported that rCBF measured by a spin-labeling method using MRI had a linear relationship with arterial pCO_2 in anesthetized rats and that the value of $\Delta CBF/\Delta pCO_2$ was 0.052 [25]. Despite variations in species, anesthetic conditions and measurement methods, our data demonstrating the linear relationship

between rCBF and pCO_2 in rats are consistent with these previous studies.

Our study has several implications for future research in neurophysiology. First, the current method could be used for repeated measurements of quantitative CBF in a single experiment. This could be useful, for example, to evaluate acute hemodynamic changes after pharmacological intervention or during behavioral tasks. It also allows measurement of the cerebral metabolic rate of oxygen utilization, which requires different PET scans using multiple tracers, such as ^{15}O-labeled water, oxygen and carbon monoxide [26]. Second, the current AV-shunt method could be used for other radiotracers not only for PET but also for single photon emission computed tomography since the GSO detector we used is sensitive to a wide range of photon energies [7]. Third, the current AV-shunt technique can be used in the future to validate more empirical methods of image-based TAC derivation (e.g., Yee et al. [2]) under various possible conditions.

There are several limitations in the AV-shunt technique. First, this method may be technically demanding and particularly difficult when measuring very frequently on different days in the same animal. The catheterized femoral artery and vein were sutured and occluded after the experiment; thus, it is difficult to repeat this study more than twice in the same animal. This could be overcome by using microsurgery to suture the arterial walls without occluding the lumen. Second, the estimation of dispersion may be difficult in each measurement due to the statistical noise in the measured TAC. Therefore, preparation of the AV shunt must be performed in the same manner, particularly in terms of the length and caliber of the shunt tube, site of artery catheter placement and physiological conditions of the animals, which may affect the laminar flow of sampling blood.

5. Conclusions

We confirmed the stability of the AV-shunt technique for measurement of the arterial input function in small animals. The AV-shunt method allowed us to obtain input TAC with less statistical noise than the image-based one and to calculate rCBF using the micro-PET scanner, and proved to be quantitative in identifying physiological changes in rCBF. In contrast, the image-based TAC suffered from the partial volume and could not be fully recovered even after correction particularly during the initial peak phase of input TAC of $[^{15}O]H_2O$. The AV-shunt technique could be useful for less invasive measurement of cerebral hemodynamics in small animals than the conventional blood sampling method.

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The use of magnetic resonance cell tracking to monitor endothelial progenitor cells in a rat hindlimb ischemic model

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ABSTRACT

A water-soluble magnetic resonance imaging (MRI) contrast agent, Dextran mono-N-succinimidyl 1,4,7,10-tetraazacyclododecane-1,4,7,10-tetraacetate-gadolinium^{3+} (Dex-DOTA-Gd^{3+}), was shown to enable monitoring of the anatomical migration and the survival period of transplanted stem cells for up to 1 month. Gadolinium molecules in the cells were rapidly eliminated from the site and excreted upon cell death. Endothelial progenitor cells (EPCs) transplanted into the inguinal femoral muscle of rats migrated distally through the knee in rats after hindlimb ischemia but did not migrate in non-ischemic rats. Interestingly, the survival period of transplanted EPCs was notably prolonged in the ischemic limb, indicating that EPCs are required by the ischemic tissues and that the fate of transplanted EPCs was affected by the disease. Compared to the commonly used particle type of MRI contrast agents, the system described in this study is expected to be invaluable to help clarifying the process of stem cell transplantation therapy.

1. Introduction

In the last decades, transplantation of hematopoietic stem cells, bone marrow (BM)-derived mesenchymal stem cells, and adipose tissue-derived mesenchymal stem cells has been reported to stimulate regeneration of different tissues such as ischemic limb [1–6], ischemic myocardium [7–9], and brain [10]. Endothelial progenitor cells (EPCs), which are crucial in the regeneration of injured tissues and organs, act through a molecular mechanism known as angiogenesis [10,11]. This post-natal revascularization starts with the recruitment of EPCs from stem cell reservoirs such as BM [12,13]. According to the paradigm of angiogenesis, EPC transplantation may offer a source of feasible endothelial cells that migrate and proliferate from preexisting blood vessels, thus promoting revascularization in the ischemic limb, contributing to angiogenesis [10,14]. Although the mechanisms of homing and differentiation of EPCs are not well understood, tissue ischemia results in the upregulation of several angiogenic factors such as vascular endothelial growth factor (VEGF), which interacts with 2 tyrosine kinase receptors, VEGF receptor-2 (VEGF-R2 or fetal liver kinase 1 [Flk-1], kinase domain receptor [KDR]) and VEGF receptor-1 (VEGFR1, Flt-1), stromal cell-derived factor-1 (SDF-1), insulin-like growth factor-1 (IGF-1), and hepatocyte growth factor (HGF). These growth factors are all expressed in several cell types, including EPCs, that promote migration to the injury site [15,19]. However, some studies have shown that neovascularization of ischemic tissues may not be exclusively attributed to the incorporation of EPCs into the vessels, but also to paracrine effects that probably influence the process of angiogenesis by releasing pro-angiogenic factors [20,21].

The ability to noninvasively track cells in vivo and follow their location after transplantation would significantly contribute to our understanding of the effects of stem cell therapy. Magnetic resonance imaging (MRI) is one of the noninvasive methods employed to examine the migratory behavior of stem and progenitor cells based on its whole-body spatial resolution, even in three-dimensional images. In MRI cell tracking, contrast agents are used to improve the quality of an image or to label transplanted cells to facilitate the detection of their location within the tissue. The most commonly used contrast agents, which have been successfully employed in tracking a variety of cell lineages, are superparamagnetic iron oxide (SPIO) and its derivatives. Progenitor cells have been tracked in vitro by using Tat peptide-derived SPIO. The high contrast of SPIO-labeled cells allows imaging of a single cell [22]. Other studies have demonstrated the feasibility of imaging

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