Published in final edited form as:

Magn Reson Imaging. 2017 December ; 44: 119–124. doi:10.1016/j.mri.2017.08.011.

Reduced distortion artifact whole brain CBF mapping using blip-reversed non-segmented 3D echo planar imaging with pseudo-continuous arterial spin labeling

Neville D. Gai^{a,*}, Yi Yu Chou^{a,b}, Dzung Pham^{a,b}, and John A. Butman^a

^aRadiology and Imaging Sciences, Clinical Center, National Institutes of Health, Bethesda, MD, USA

^bCenter for Neuroscience and Regenerative Medicine, Henry M. Jackson Foundation, Bethesda, MD, USA

Abstract

Purpose—To implement and evaluate interleaved blip-up, blip-down, non-segmented 3D echo planar imaging (EPI) with pseudo-continuous arterial spin labeling (pCASL) and post-processing for reduced susceptibility artifact cerebral blood flow (CBF) maps.

Materials and methods—3D EPI non-segmented acquisition with a pCASL labeling sequence was modified to include alternating k-space coverage along phase encoding direction (referred to as “blip-reversed”) for alternating dynamic acquisitions of control and label pairs. Eight volunteers were imaged on a 3T scanner. Images were corrected for distortion using spatial shifting transformation of the underlying field map. CBF maps were calculated and compared with maps obtained without blip reversal using matching gray matter (GM) images from a high resolution 3D scan. Additional benefit of using the correction for alternating blip-up and blip-down acquisitions was assessed by comparing to corrected blip-up only and corrected blip-down only CBF maps. Matched Student t-test of overlapping voxels for the eight volunteers was done to ascertain statistical improvement in distortion.

Results—Mean CBF value in GM for the eight volunteers from distortion corrected CBF maps was 50.8 ± 9.9 ml/min/100 gm tissue. Corrected CBF maps had 6.3% and 4.1% more voxels in GM when compared with uncorrected blip up (BU) and blip down (BD) images, respectively. Student t-test showed significant reduction in distortion when compared with blip-up images and blip-down images ($p < 0.001$). When compared with corrected BU and corrected BD only CBF maps, BU and BD corrected maps had 2.3% and 1% more voxels ($p=0.006$ and 0.04 , respectively).

Conclusion—Pseudo-continuous arterial spin labeling with non-segmented 3D EPI acquisition using alternating blip-reversed k-space traversal and distortion correction provided significantly better matching GM CBF maps. In addition, employing alternating blip-reversed acquisitions during pCASL acquisition resulted in statistically significant improvement over corrected blip-up and blip-down CBF maps.

*Corresponding author at: Radiology & Imaging Sciences, National Institutes of Health, Rm 1C502, 9000 Rockville Pike, Bethesda, MD 20892, USA. Neville.Gai@nih.gov (N.D. Gai).

Keywords

3D pCASL; Arterial spin labeling; 3D EPI, distortion correction; CBF maps; Blip-up blip-down

1. Introduction

Arterial spin labeling (ASL) [1] provides non-invasive measurements of cerebral blood flow by using labeled blood as an endogenous contrast agent. Measurement of cerebral perfusion can provide unique information not achievable with other MRI contrasts in a number of clinical situations, e.g. differentiation of tumor recurrence from radiation necrosis.

Various techniques have been proposed for arterial spin labeling. A recent consensus review on the subject recommended the use of pseudo-continuous arterial spin labeling (pCASL) due to its superior labeling efficiency and SNR as well as compatibility with RF transmission hardware [2]. For the readout module of ASL, segmented 3D acquisition is preferred due to SNR efficiency and relative insensitivity to off-resonance effects. However, non-segmented acquisition offers relatively higher SNR efficiency and reduced variation in signal arising from phase inconsistencies in multi-shot techniques. Such techniques include 3D GRAdient and Spin Echo (GRASE) acquisition with Flow sensitive Alternating Inversion Recovery (FAIR) pulsed arterial spin labeling [3] and with CASL [4]. The use of single-shot 3D fast spin-echo spiral acquisitions has also been demonstrated [5]. While fast spin-echo and GRASE acquisitions have the benefit of reducing T2* related artifacts that occur with gradient-echo EPI acquisitions, this comes at the expense of higher RF power deposition and prolonged acquisition times relative to pure EPI readouts. Non-segmented 3D EPI acquisition with CASL [6] and pulsed ASL [7] has been demonstrated using centric-ordered 3D Echo Planar Imaging which provides high k-space sampling efficiency allowing for full brain coverage with high SNR.

One drawback with ns-3D EPI acquisition is image distortion resulting from static field inhomogeneity and susceptibility. This distortion can be reduced by minimizing echo time. Further reduction in distortion can be achieved by use of distortion correction techniques. Of the several distortion correction techniques that have been proposed, “blip reversal” provides a simple yet effective means for distortion correction without the need to explicitly acquire field inhomogeneity maps [8]. The distorted images are processed to correct for pixel shift as well as magnitude changes. This idea has been applied to EPI by traversing k-space in opposite directions [9]. A robust least squares solution for reconstructing the undistorted image from a displacement field and two distorted images with opposing polarity has been validated for spin-echo EPI images [10]. It has also been successfully applied to DWI using 3D EPI acquisition [11]. This distortion correction scheme is particularly well suited for arterial spin labeling acquisition since multiple pairs of control and labeled images are acquired, thereby circumventing any increase in scan time.

Madai et al. [12] employed distortion correction for 3D GRASE acquisition with pulsed arterial labeling where a separate pair of opposing phase-encoded M0 images (images with no tagging or preparation pulses) was acquired to correct for the unidirectional phase encoded ASL acquisition.

In this work, we acquired k-space reversed ns-3D EPI for alternating pairs of control and label image sets. The distortion was derived from the mean of these pairs and applied to the acquired bidirectional phase encoded ASL images.

2. Materials and methods

2.1. Sequence

Non-segmented 3D EPI acquisition with pCASL was modified so that blip-reversed images were acquired in alternate dynamic scans. Accordingly, the first set of tagged and control images (first dynamic scan) were acquired with k_y space traversed from $-k_y$ to $+k_y$ (logical blip-up) while a second pair (second dynamic scan) was acquired with k_y space acquisition from $+k_y$ to $-k_y$ (logical blip-down). This was achieved by reversing the phase encoding blips. The pattern was maintained till the last dynamic pair. To further reduce echo time, phase encoding was chosen to correspond to the smaller FOV direction. For transaxial acquisition, this corresponded with the physical left to right direction.

2.2. Scanning

Eight volunteers (5M, 3F; age: 42.5 ± 6.9 yrs) were scanned on a 3T Philips Achieva TX scanner (Philips Medical Systems, Best, The Netherlands) using an IRB approved protocol and after informed consent was obtained. An 8 channel receive only head coil was employed for acquisition. The pCASL labeling sequence used had the following parameters: label duration = 1.65 s, label delay = 1.8 s; 3D EPI acquisition parameters were: FOV = $24 \text{ cm} \times 20 \text{ cm}$, resolution: $3 \times 3 \times 4 \text{ mm}^3$, EPI TR/TE = 22/11 ms, centric k_z encoding, SENSE (y) = 2.5, 30 slices, phase encoding (L/R), spectral-spatial excitation with a 1-2-1 binomial pulse for fat suppression ($\theta = 25^\circ$), 34 dynamics, scan time: 4 min 50 s. To reduce blurring resulting from extended T1 relaxation, an optimal flip angle train was employed [7,13]. A background suppression (BS) adiabatic inversion pulse was used at TI = 1.8 s after the initial saturation pulse used to saturate magnetization in the imaging slab. In Bloch simulations, using literature T1 values for gray matter (GM), white matter (WM) and cerebrospinal fluid (CSF) [14,15], the inversion pulse suppressed the signal from the three tissue types. Normalized magnetization was 0.85, 0.96 and 0.52 in static GM, WM and CSF, respectively, without this inversion pulse and was 0.37, 0.62 and 0.09 with the inversion pulse.

2.3. Postprocessing

2.3.1. Distortion correction—The post-processing pipeline included the following steps: (a) 3D rigid registration of all images to the first acquired blip-up (BU) or blip-down (BD) control image was performed using FLIRT [16] to correct for any motion. (b) The susceptibility-induced off resonance field was estimated based on the average (higher SNR) of all blip-up and blip-down control images using the “top-up” algorithm [17]. This field allowed undistorted images to be computed for all label and control image pairs. (c) All labeled and control distortion corrected images were averaged to obtain 2 sets of images. (d) Finally, the control images were subtracted from the labeled images to get reduced distortion cerebral blood flow (CBF) images.

Steps (b) through (d) were also applied to only the BU or BD image sets separately in order to ascertain reduced distortion resulting from using correction on both BU and BD sets.

2.3.2. Registration and GM mask—To ascertain reduction in distortion, a mask was prepared for each slice from relatively distortion free high resolution (1 mm³) 3D magnetization prepared rapid gradient echo (MPRAGE) images and compared with the CBF images in the following manner: (a) The skull was stripped using SPECTRE [18] (b) GM was segmented from other tissues using MALP-EM [19]. (c) The MPRAGE images were registered (rigid only) to the BU, BD and distortion corrected CBF images using FLIRT. (d) The number of voxels which overlapped in matched GM from MPRAGE and CBF images from the three sets (BU, BD and distortion corrected) was counted over the entire brain for all eight volunteers. Steps (c) and (d) were also repeated with corrected BU only and corrected BD only CBF images.

2.3.3. CBF map—CBF values were calculated using the formulation from Alsop et al. [20]

$$CBF = \frac{\lambda \Delta M \exp\left(\frac{T_{art}}{T_{1a}}\right)}{2M_{0B} T_{1br} \left[\exp\left(\frac{\min(T_{art} - PLD0)}{T_{1br}}\right) - \exp\left(-\frac{PLD}{T_{1br}}\right) \left(1 - \frac{T_{1RF}}{T_{1br}}\right) \right]}$$

where λ is the blood brain partition coefficient, M is the difference signal from control and label images, T_{art} refers to the arterial transit time, T_{1a} is the T₁ of arterial blood, α is the inversion efficiency, M_{0B} is the equilibrium brain tissue magnetization, T_{1br} is the T₁ of brain tissue, PLD is the post-labeling delay and T_{1RF} refers to T₁ of brain tissue in the presence of off-resonance irradiation. $PLD = 1800$ ms while other constants were as defined in [21]. To account for the inversion pulse at $TI_1 = 1800$ ms, equilibrium M_{0B} was derived from measured $M_{0B,m}$ in the control image using

$$M_{0B} = \frac{M_{0B,m}}{\left[1 - 2 \exp\left(\frac{-TI_2}{T_{1,br}}\right) + \exp\left(\frac{-TI_1 - TI_2}{T_{1,br}}\right) \right]}$$

where $TI_2 = 1650$ ms. Mean CBF values in GM were calculated over all 30 slices for BU, BD and distortion corrected images using the GM mask generated from high resolution 3D MPRAGE images.

Student's *t*-test (matched, 1-tailed) was done to check for significant increases between the voxels overlapping the GM mask for the BU, BD and distortion corrected CBF images. A similar test was performed to ascertain significant increases in GM overlap for the corrected BU only and BD only images in relation to the corrected BU-BD data set.

In addition, a paired *t*-test was also performed to ascertain differences in mean GM CBF values between BU, BD and distortion corrected CBF maps.

3. Results

Fig. 1 shows the comparison between distortion corrected images and the uncorrected images. In the control images, the distortion along the phase encoding direction (R/L) is clearly evident, particularly in the frontal lobes and ventricles. A similar distortion is seen in CBF images. The asymmetry of the frontal horns and the signal pileup in the frontal white matter is corrected by application of the distortion correction algorithm.

The bottom row shows the overlap of the cortical gray matter mask with the thresholded perfusion map. Ideally, all voxels in the gray matter mask should correspond to highly perfused tissue. Most areas of improvement were in the superior and medial frontal gyrus.

Fig. 2 shows the bar chart of the matching CBF GM voxels in the eight volunteers for (a) BU (b) BD and (c) corrected CBF maps. On average, there were 6.3% more voxels matching with the MPRAGE prepared mask in the distortion corrected images when compared with BU and 4.1% more when compared to BD images. Pairwise comparison of voxels matching GM for BU and BD images with distortion corrected images showed significant differences ($p = 0.00007$ and $p = 0.00009$, respectively) indicating significantly improved localization of the CBF signal.

Fig. 3 shows the bar chart of matching CBF GM voxels for the case of corrected BU, corrected BD and BU-BD corrected maps. Corrected BU-BD CBF GM showed a mean 2.3% increase in overlap over corrected BU only CBF GM with a MPRAGE prepared GM mask. This increase was 1% when compared with corrected BD only CBF GM. Both increases were significant in a paired t -test ($p = 0.006$ and $p = 0.04$, respectively). Fig. 4 provides an example of the improvement in distortion correction seen when both BU and BD images were acquired during pCASL acquisition and corrected as opposed to performing distortion correction to unidirectional phase encoded pCASL data acquisition (corrected BU or BD only).

Fig. 5 shows six slices (every fifth slice of 30) of distortion corrected CBF images in a volunteer. Mean GM CBF values across the eight volunteers for blip-up, blip-down and corrected maps were 50.9 ± 7.8 , 54.7 ± 10.7 and 50.8 ± 9.9 ml/min/100 g, respectively. There was no significant difference in the mean CBF GM values (BU vs corrected, $p = 0.98$ and BD vs corrected, $p = 0.51$).

4. Discussion

Non-segmented 3D EPI with ramp sampling potentially offers relatively higher SNR per unit time when compared with other 3D Cartesian (GRASE) and non-Cartesian sampling techniques due to almost continuous sampling of k-space. In addition, when combined with widely available parallel imaging techniques, the entire k-space corresponding to full brain coverage (with typically employed resolution for ASL) can be covered in an efficient manner. Non-segmented acquisition also ameliorates phase related inconsistencies encountered with segmented EPI acquisitions. With centric encoding, the need for multiple delay times used in 2D multiple slice encodings is also avoided providing better CBF estimates over the entire volume. A longer readout can result in T1 relaxation related

blurring which can be overcome using a modulated flip angle scheme as done here. A drawback in comparison to SE based techniques and GRASE acquisition is increased distortion due to field inhomogeneity and susceptibility. Therefore, a better case can be made for using distortion correction with 3D ns-EPI acquisition due to increased distortion over 3D GRASE [12]. In addition, based on an earlier work on distortion correction for DTI [22], the use of interleaved blip-up, blip-down EPI acquisitions during ASL acquisition leads to a more accurate map (and correction) with reduced effect from motion, magnetic field drift and shimming related discrepancies. In addition, as demonstrated in this work, application of the field map to bidirectional phase encoded images may result in a more accurate correction as areas of signal pile-up may be better unraveled by corresponding areas of signal stretching. Despite a 1% increase in matching voxels when compared to corrected BD only derived GM CBF map, in practice, one cannot predetermine which of the two unidirectional phase encoded sets should be acquired. Hence, the worst case scenario might hold so that the greater of the two errors is equally likely.

The sequence provides sufficient perfusion signal to achieve whole brain coverage in a time efficient manner. No additional scan time is required when compared with standard ns-3D EPI acquisition [7] since a number of dynamics are always employed for ASL. Phase encoding was chosen to be along R/L since $FOV_y < FOV_x$, thereby reducing the phase encoding steps and echo time (and therefore distortion) even prior to correction. By interleaving the BU-BD acquisitions, a temporally averaged map of the distortion field was measured. Susceptibility effects with a temporal variation over the scan time would therefore be averaged out with the current implementation. Spectral-spatial RF excitation pulse was favored over a fat saturation (FS) pulse since residual fat signal shifted into the brain (when FS is used) made distortion correction difficult. In addition, susceptibility induced signal loss in the through-plane direction as well as in areas of strong field offset with gradient echo EPI will lead to signal dropout and cannot be restored. This is noticeable in the areas close to the orbits and cavernous sinus.

Typically, two or more BS pulses are employed with pCASL to suppress background signal. Each BS pulse leads to a reduction in the pCASL perfusion signal. Since our goal here was the implementation of a distortion correction technique, use of BS pulses was peripheral to the implementation and was primarily employed to suppress pulsatile CSF signal which can contaminate CBF maps.

The number of volunteers was limited in our study and was used for proof of concept and for limited statistical analysis. Although mean GM CBF values were not significantly different between BU, BD and corrected images, differences were mainly a result of measured M_{0B} ($M_{0B,m}$) in the three images types. A separate equilibrium M_{0B} image could have been obtained but we chose to measure M_{0B} in the unprocessed BU, BD and corrected images as done in [21]. Measuring GM signal in an ROI on lower resolution EPI images can result in considerable variability due to partial volume effect even in deep cerebri and cerebellar gray matter. Absolute quantification of CBF in GM is also limited by the assumed value for the parameters α , λ , T_{1br} , T_{1RF} and T_{art} . Nevertheless, our mean GM CBF of 50.8 ml/min/100 g tissue was in agreement with values reported in prior studies [23,24]. Local GM CBF values in areas of distortion corrected by the BU-BD technique would show larger

disparities. Such regional analysis was not carried out in this work since geometric distortion analysis showed clear evidence of improvement.

In conclusion, blip-reversed ns-3D EPI acquisition along with post-processing for distortion correction significantly reduced GM misregistration in CBF maps without increasing scan time. In addition, by incorporating blip-reversed acquisition in an interleaved fashion during the pCASL acquisition, better distortion correction may be obtained when compared with a technique employing unidirectional phase encoding but relying on separate phase reversed equilibrium M_{0B} image sets to determine distortion correction.

Abbreviations

ns-3D EPI	non-segmented 3D echo planar imaging
pCASL	pseudo-continuous arterial spin labeling
CBF	cerebral blood flow
GRASE	GRadient and Spin Echo
FAIR	Flow sensitive Alternating Inversion Recovery
BU	blip up
BD	blip down
FLIRT	FLexible Image Registration Toolbox
SPECTRE	Simple Paradigm for Extra-Cerebral Tissue Removal
MALP-EM	Multi-Atlas Label Propagation with Expectation-Maximization based refinement
MPRAGE	Magnetization Prepared Rapid Gradient Echo
PLD	Post-labeling delay
λ	blood brain partition coefficient
T_{art}	arterial transit time
T_{1a}	T1 of arterial blood
α	inversion efficiency
M_{0B}	equilibrium brain tissue magnetization
T_{1br}	T1 of brain tissue
T_{1RF}	T1 of brain tissue in the presence of off-resonance irradiation
R/L	right-to-left

References

1. Williams DS, Detre JA, Leigh JS, Koretsky AP. Magnetic resonance imaging of perfusion using spin inversion of arterial water. *Proc Natl Acad Sci U S A*. 1992; 89(1):212–6. [PubMed: 1729691]
2. Alsop DC, Detre JA, Golay X, et al. Recommended implementation of arterial spin-labeled perfusion MRI for clinical applications: a consensus of the ISMRM perfusion study group and the European consortium for ASL in dementia. *Magn Reson Med*. 2015; 73(1):102–16. [PubMed: 24715426]
3. Gunther M, Oshio K, Feinberg DA. Single-shot 3D imaging techniques improve arterial spin labeling perfusion measurements. *Magn Reson Med*. 2005; 54(2):491–8. [PubMed: 16032686]
4. Fernandez-Seara MA, Wang Z, Wang J, et al. Continuous arterial spin labeling perfusion measurements using single shot 3D GRASE at 3 T. *Magn Reson Med*. 2005; 54(5):1241–7. [PubMed: 16193469]
5. Duhamel, G., Alsop, DC. Proceedings of the 12th annual meeting of ISMRM. Kyoto, Japan: 2004. Single-shot susceptibility insensitive whole brain 3D fMRI with ASL; p. 518
6. Talagala, SL., Slavin, GS., Ostuni, J., Chesnick, S. Proceedings of the 14th annual meeting of the Society of Magnetic Resonance. Seattle, Washington, USA: 2006. CASL perfusion MRI with non-segmented low flip angle 3D EPI; p. 3422
7. Gai ND, Talagala SL, Butman JA. Whole-brain cerebral blood flow mapping using 3D echo planar imaging and pulsed arterial tagging. *J Magn Reson Imaging*. 2011; 33(2):287–95. [PubMed: 21274969]
8. Chang H, Fitzpatrick JMA. Technique for accurate magnetic resonance imaging in the presence of field inhomogeneities. *IEEE Trans Med Imaging*. 1992; 11(3):319–29. [PubMed: 18222873]
9. Morgan PS, Bowtell RW, McIntyre DJ, Worthington BS. Correction of spatial distortion in EPI due to inhomogeneous static magnetic fields using the reversed gradient method. *J Magn Reson Imaging*. 2004; 19(4):499–507. [PubMed: 15065175]
10. Andersson JL, Skare S, Ashburner J. How to correct susceptibility distortions in spin-echo echo-planar images: application to diffusion tensor imaging. *Neuroimage*. 2003; 20(2):870–88. [PubMed: 14568458]
11. Gallichan D, Andersson JL, Jenkinson M, Robson MD, Miller KL. Reducing distortions in diffusion-weighted echo planar imaging with a dual-echo blip-reversed sequence. *Magn Reson Med*. 2010; 64(2):382–90. [PubMed: 20665782]
12. Madai VI, Martin SZ, von Samson-Himmelstjerna FC, et al. Correction for susceptibility distortions increases the performance of arterial spin labeling in patients with cerebrovascular disease. *J Neuroimaging*. 2016; 26(4):436–44. [PubMed: 26902457]
13. Stehling MK. Improved signal in “snapshot” FLASH by variable flip angles. *Magn Reson Imaging*. 1992; 10(1):165–7. [PubMed: 1545677]
14. Stanisz GJ, Odobina EE, Pun J, et al. T1, T2 relaxation and magnetization transfer in tissue at 3T. *Magn Reson Med*. 2005; 54(3):507–12. [PubMed: 16086319]
15. Gai ND, Butman JA. Modulated repetition time look-locker (MORTLL): a method for rapid high resolution three-dimensional T1 mapping. *J Magn Reson Imaging*. 2009; 30(3):640–8. [PubMed: 19630081]
16. Modersitzki J. FLIRT with rigidity: image registration with a local non-rigidity penalty. *Int J Comput Vis*. 2008; 76:153–63.
17. Smith SM, Jenkinson M, Woolrich MW, et al. Advances in functional and structural MR image analysis and implementation as FSL. *Neuroimage*. 2004; 23(Suppl 1):S208–19. [PubMed: 15501092]
18. Carass A, Cuzzocreo J, Wheeler MB, Bazin PL, Resnick SM, Prince JL. Simple paradigm for extra-cerebral tissue removal: algorithm and analysis. *Neuroimage*. 2011; 56(4):1982–92. [PubMed: 21458576]
19. Ledig C, Heckemann RA, Hammers A, et al. Robust whole-brain segmentation: application to traumatic brain injury. *Med Image Anal*. 2015; 21(1):40–58. [PubMed: 25596765]

20. Alsop DC, Detre JA. Reduced transit-time sensitivity in noninvasive magnetic resonance imaging of human cerebral blood flow. *J Cereb Blood Flow Metab.* 1996; 16(6):1236–49. [PubMed: 8898697]
21. Aslan S, Xu F, Wang PL, et al. Estimation of labeling efficiency in pseudocontinuous arterial spin labeling. *Magn Reson Med.* 2010; 63(3):765–71. [PubMed: 20187183]
22. Irfanoglu MO, Modi P, Nayak A, Hutchinson EB, Sarlls J, Pierpaoli C. DR-BUDDI (diffeomorphic registration for blip-up blip-down diffusion imaging) method for correcting echo planar imaging distortions. *Neuroimage.* 2015; 106:284–99. [PubMed: 25433212]
23. Pantano P, Baron JC, Lebrun-Grandie P, Duquesnoy N, Bousser MG, Comar D. Regional cerebral blood flow and oxygen consumption in human aging. *Stroke.* 1984; 15(4):635–41. [PubMed: 6611613]
24. Zhang K, Herzog H, Mauler J, et al. Comparison of cerebral blood flow acquired by simultaneous [15O]water positron emission tomography and arterial spin labeling magnetic resonance imaging. *J Cereb Blood Flow Metab.* 2014; 34(8):1373–80. [PubMed: 24849665]

Fig. 1.

Distortion correction demonstrated in a CBF map where the ASL image is overlaid on a gray matter mask obtained from high resolution relatively distortion free 3D MPRAGE sequence. Row 1 shows the control image, row 2 shows the CBF map while row 3 shows a thresholded CBF map overlaid on GM mask. Columns 1, 2 and 3 correspond to blip-down EPI (physical $R \rightarrow L$ encoding), blip-up EPI and distortion corrected case, respectively.

Fig. 2.
Bar chart of matching GM voxels from CBF maps obtained from BU, BD and distortion corrected pCASL 3D EPI images in eight volunteers.

Fig. 3.
Bar chart of matching GM voxels from CBF maps obtained from corrected BU only, corrected BD only and distortion corrected pCASL 3D EPI images in eight volunteers.

Fig. 4.

Improvement in distortion correction when using bidirectional phase encoded acquisition with pCASL as opposed to unidirectional phase encoded data acquisition with pCASL demonstrated with CBF maps showing regions where the GM mask from 3D MPRAGE does not overlap the ASL image. Column 1, 2 and 3 correspond to corrected blip-down only, corrected blip-up only and corrected BU-BD CBF maps.

Fig. 5. CBF map of every fifth slice from the 3D set obtained after distortion correction with the 3D non-segmented BU-BD EPI pCASL technique described. CBF values are in ml/min/100 g of tissue.