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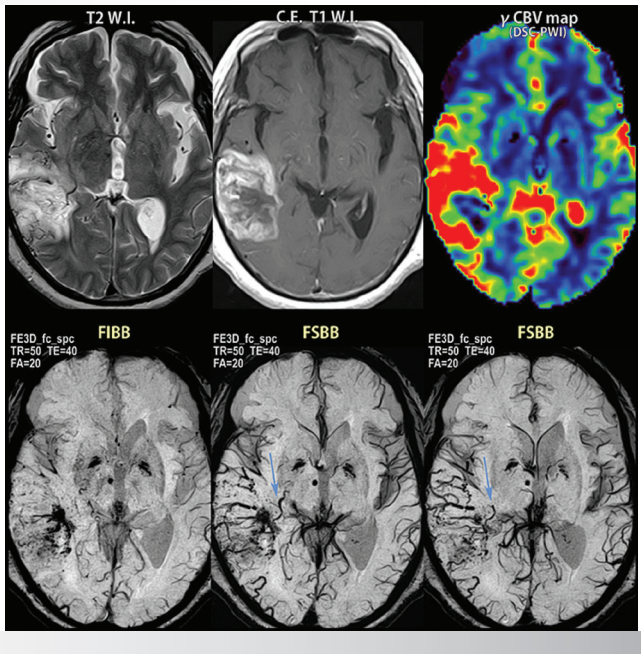
FSBB:
Flow Sensitive Black Blood Imaging



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Toshiba's Flow-Sensitive Black Blood (FSBB) sequence applies motion probing gradients (MPGs) to T2*-weighted sequences to generate significantly improved vessel contrast without suffering from excessive T2* decay [1], [2].

Introduction

Time of Flight (TOF) is the most widely used Magnetic Resonance Angiography (MRA) technique, largely in part to the simplicity of its implementation. The TOF sequence is an inflow technique by which the volume of interest is repeatedly excited in order to saturate background signal. Inflowing blood originating outside of the volume is unsaturated, resulting in higher signal intensity relative to saturated background tissue, thus generating a bright blood image. The downside to the simplicity of TOF is that the bright blood contrast mechanism requires fast coherent flow, limiting the amount and type of flow that can be visualized. To visualize slow or turbulent blood flow, such as in the veins, T2*-weighted sequences sensitive to susceptibility differences have been used because of the higher susceptibility of de-oxygenated hemoglobin in venous blood. While T2*-weighted imaging can help visualize slower flowing vessels, the contrast is limited as the overall image quality deteriorates with increased T2* decay. To overcome this limitation, Toshiba's Flow-Sensitive Black Blood (FSBB) sequence applies motion probing gradients (MPGs) to T2*-weighted sequences to generate significantly improved vessel contrast without suffering from excessive T2* decay [1], [2].

Background

Conventional clinical MR contrast is generated by manipulating the repetition and echo times (TR and TE respectively) to yield images with various T1 and T2 weightings. T1 is the longitudinal relaxation time associated with the recovery of the magnetization back to the equilibrium +z orientation, and T2 is the transverse relaxation time associated with the decay of the observable signal. In general, both of these relaxation times are primarily influenced by the molecular motion of local water molecules and are insensitive to gradients created by magnetic field inhomogeneity. Magnetic field inhomogeneity on the macroscopic length scale results in a net phase change in standard field echo (FE) sequences and generally complicates things like chemical-shift selective fat suppression. On the other hand, microscopic magnetic field inhomogeneity produces a phase distribution of the magnetization on the sub-voxel level that results in a net signal reduction, called T2* decay. Unlike T2 decay, signal loss from T2* can be recovered with the use of spin echoes and can only be observed with FE sequences.

Susceptibility

Magnetic field inhomogeneity originates from differences in susceptibility between tissues. Susceptibility is a measure of how efficiently a material will respond to an applied magnetic field. Subsequently, at the interface of two materials with different magnetic susceptibility, there will be a distortion of the magnetic field, resulting in an inhomogeneous magnetic field. Macroscopically, this effect is most commonly observed at air-tissue interfaces and is particularly problematic in regions such as the brachial plexus and sinuses. When oxygenated, hemoglobin is diamagnetic and does not result in a significant magnetic field distortion. However, hemoglobin becomes paramagnetic when deoxygenated and therefore results in a local magnetic field distortion that generates a phase distribution on the sub-voxel dimension scale. The phase distribution within the voxel results in a net signal reduction in the image, called T2* decay. As a result, T2* sequences are particularly sensitive to microscopic susceptibility differences and are useful in imaging venous blood, hemorrhage, and iron deposits. The difficulty with using longer TEs in T2* sequences is that the overall image quality begins to suffer, greatly limiting the extent by which small differences in susceptibility can be reliably visualized.

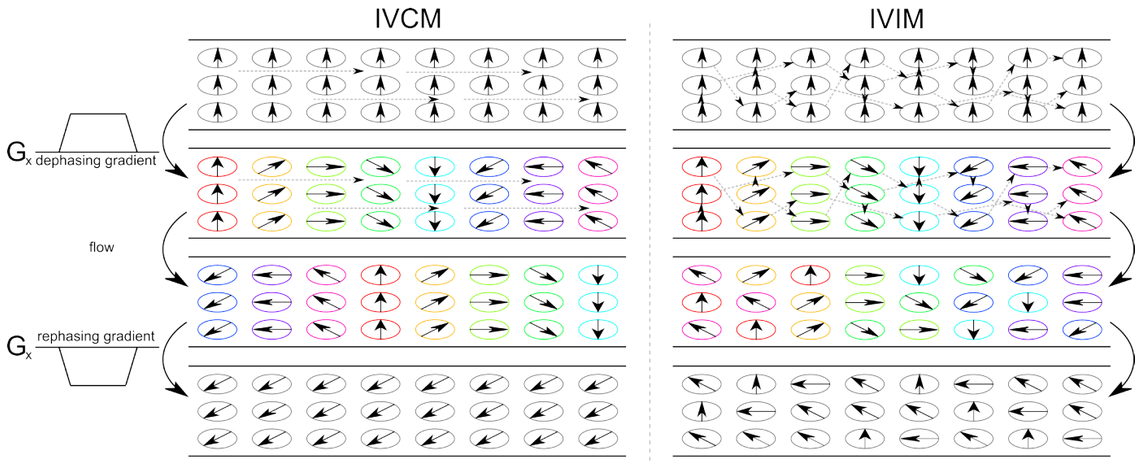

Figure 1 – Comparison of intra-voxel coherent motion (IVCM, left) and intra-voxel incoherent motion (IVIM, right) in the presence of motion-probing gradients (MPGs). In both cases, the net flow is going to the right. The first MPG along the x-direction dephases the net magnetization by creating a distribution of phases indicated by the different colors in the second row. In the third row, with IVCM, all the spins move coherently allowing them to maintain their phase distribution. With IVIM, the phase distribution of all the spins becomes mixed up because the spins do not move coherently. Upon reversal of the dephasing gradient along the x-direction (fourth row), all of the spins are rephased in the case of IVCM and the signal is restored to its original intensity with an added phase change. With IVIM, there is incomplete rephasing resulting in a reduced net signal amplitude.

To remedy this limitation, Toshiba's FSBB sequence applies a weak motion probing gradient to sensitize the magnetization to very slow flow.

Motion Probing Gradients

In Diffusion Weighted Imaging (DWI), bipolar gradients are used to sensitize the magnetization to molecular motion. The length-scale being probed is often described by the b-value, such that:

$$b = -\gamma^2 G^2 \delta^2 \left(\Delta - \frac{\delta}{3}\right),$$

where γ is the gyromagnetic ratio, G is the amplitude of the gradient pulse, δ is the duration of the gradient pulse, and Δ is the time between the pair of gradients [3]. As the b-value is directly proportional to the square of the gradient amplitude, the length-scale to investigate can be finely tuned by simply changing the gradient amplitude. To probe motion on the molecular length scale, strong gradients are used such that b = 1000s/mm^2. For vessel motion that occurs on a length-scale much larger than molecular diffusion, the applied gradients are much weaker such that b = 1-3s/mm^2. In other words, probing motion on shorter length scales requires stronger gradients whereas motion on longer length scales requires weaker gradients.

Incoherent vs. Coherent Motion

Two categories are used to describe the motion of spins within an individual voxel: intra-voxel coherent motion (IVCM) and intra-voxel incoherent motion (IVIM). The category of motion has an effect on the voxel signal intensity in the presence of an MPG, as illustrated in **Figure 1**.

In the case of IVCM, prior to the application of the first MPG, the flowing spins begin with coherent phase prior to the MPG. When the first MPG is applied, each spin acquires a spatially dependent phase depending on their location in the gradient field, denoted by the different colors in **Figure 1**. Since the motion is coherent, the phase distribution retains its order as the spins flow through the vessel. When the second MPG is applied, the spins within a voxel coherently rephase. Since the spins have moved locations during the MPGs due to flow, they have acquired a different phase with each MPG, resulting in a net phase change. This phase change is the basis for velocity encoding in flow studies as the net phase change will be proportional to the velocity of the coherent motion through the voxel of interest. This flow-induced phase change can be compensated for by using first-order gradient moment nulling.

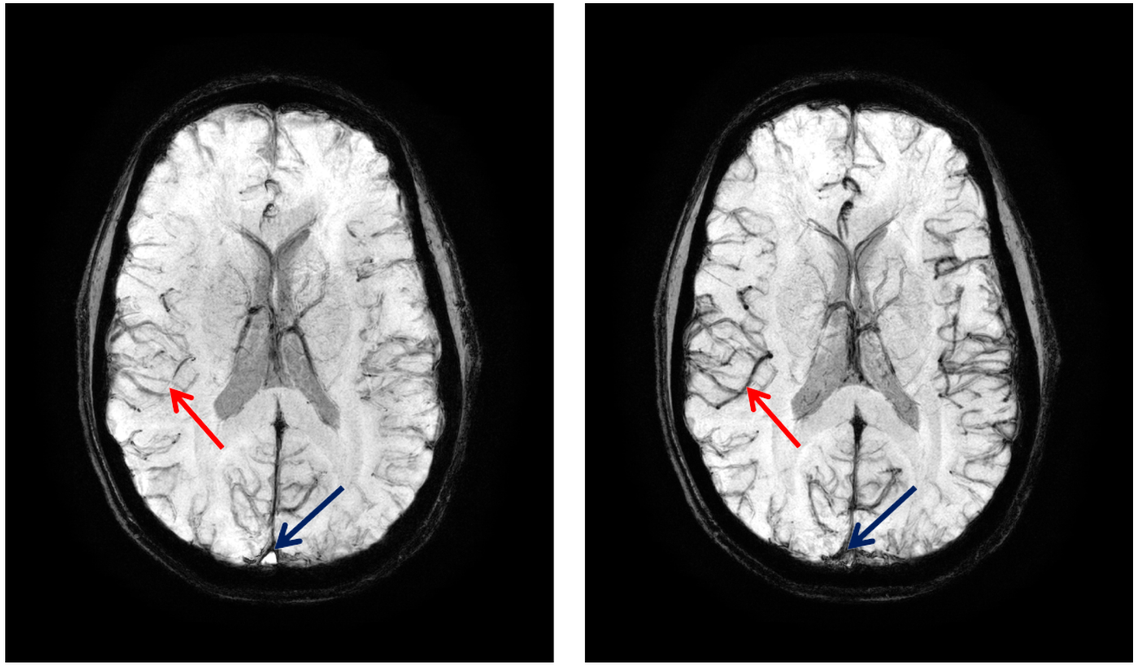

Figure 2 – Minimum intensity projection (mIP) of the magnitude images from an FIBB image (left) and an FSBB image (right) with b = 1s/mm^2. Differences in contrast due to the MPGs can be seen especially in the posterior temporal branches as marked by the red arrow and the superior sagittal sinus marked by the blue arrow.

With IVIM, prior to the application of the first MPG, the flowing spins are mixing together, but retain coherent phase since they originated with the same phase. When the first MPG is applied, each spin acquires a spatially dependent phase depending on its location within the gradient field as denoted by the different colors. Since the motion is incoherent, the different spins mix spatial locations, disturbing the ordered dephasing created by the first MPG. After some time, the second MPG with opposite polarity is applied to reverse the dephasing of the first gradient pulse. Since the ordered dephasing was disturbed by the incoherent motion, the second MPG results in incomplete rephasing of the voxel. The net effect is a distribution of spin phase within the voxel, resulting in a reduced signal intensity in locations exhibiting IVIM.

FSBB

Toshiba's Flow-Sensitive Black Blood (FSBB) technique utilizes MPGs in a 3D FE sequence to generate image contrast from both T2* susceptibility effects as well as IVIM dephasing effects. In the absence of MPGs, the FSBB sequence is referred to as a Flow-Insensitive Black Blood sequence (FIBB) that is fully flow-compensated in all three directions and generates contrast solely from T2* effects. The effect of using MPGs can be seen in **Figure 2**, comparing an FIBB image with an FSBB image. Two sets of images were acquired with identical TR/TE = 45/25ms, flip angle = 25°, FOV = 24x18cm² with 2mm slices, acquired matrix = 512x384 reconstructed to 1024x768. Minimum intensity projections (mIPs) were obtained by summing over 6 slices for a slice thickness of 12mm. The first set of images (left, FIBB), were acquired with flow-compensation, and the second set of images (right, FSBB) were acquired with MPGs with a gradient corresponding to b = 1s/mm². The FIBB images show standard T2* weighting where the slight darkening in the vessels comes solely from dephasing due to susceptibility. The FSBB images show the same T2* weighting, but with significantly improved visualization of slow-flowing vessels that were otherwise unobservable in the FIBB images. An example of a slow-flowing vessel in the posterior temporal branches that was clearly not seen in the FIBB image is pointed out by the red arrow. In the FIBB image, the signal from the super sagittal sinus has been flow-compensated but is completely dephased in the FSBB image, marked by the blue arrow.

Applications

Being sensitive to susceptibility as well as slow flowing vessels, FSBB is suitable for studies in traumatic brain injury, stroke, hemorrhage, neurodegenerative disorders, multiple sclerosis, vascular malformations, venous disease, as well as neurocysticercosis [4]. FSBB

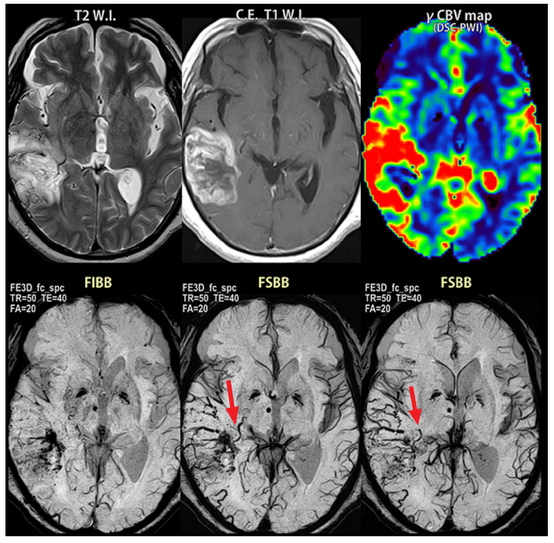

Figure 3 – Patient study of a glioma showing T2, T1, perfusion, FIBB, and two different slices from FSBB. The red arrows point to an abnormal vessel that appears to be feeding the tumor and cannot be seen in the standard FIBB image.

can also be used to aid in evaluating the characteristics and grading of tumors because of its ability to reveal vasculature that is difficult to see in other sequences. **Figure 3** shows a study of a patient with a glioma in the brain. While the tumor can be seen in both the T2- and post-contrast T1-weighted images, the images do not reveal information about the surrounding vasculature. There is a significant hypo-intensity within the tumor observable in both the perfusion weighted image (PWI) and FIBB image indicative of a hemorrhage. The increased vasculature seen in the FSBB is in agreement with the increased signal intensity surrounding the tumor in the PWI.

Other studies have found the FSBB sequence to be useful in visualizing lenticulostriate arteries (LSAs), which supply blood to the basal ganglia and its vicinity [5]. While LSAs are easily visualized in TOF methods at 7T, FSBB can visualize these vessels at the clinically available field strengths of 1.5T and 3T. The ability to better visualize LSAs has the potential application of better assessment of the risk of stroke in elderly patients [6].

Conclusion

Toshiba's Flow-Sensitive Black Blood sequence builds upon the susceptibility contrast in T2*-weighted FE sequences by adding motion probing gradients to generate additional signal dephasing in slow flowing vessels. Post-processing techniques provided by other vendors have been developed to improve vessel visibility in T2* images by using susceptibility-weighted phase masks to enhance the contrast. Being inherently a post-processing approach, the source images are still constrained by the SNR limitations of long TE T2*-weighted imaging. In addition to image quality concerns with longer TEs, the susceptibility-weighted phase masks must be filtered to compensate for background magnetic field inhomogeneity. The FSBB sequence bypasses these complications by using MPGs that allow for slow-vessel contrast to develop at shorter TEs without sacrificing imaging quality. As the MPGs create vessel contrast that is observable in unprocessed magnitude images, the FSBB sequence presents a much simpler approach that does not require specialized post-processing routines. As a result, FSBB is a valuable sequence that would be of great benefit to any neuroradiology protocol studying traumatic brain injury, stroke, hemorrhage, neurodegenerative disorders, multiple sclerosis, vascular malformations, venous disease, neurocysticercosis, and brain tumors.

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