Perfusion Tensor Imaging

L. R. Frank[1,2], E. C. Wong[1]

[1]Radiology, UC San Diego, La Jolla, CA, United States, [2]VA San Diego, La Jolla, CA, United States

Introduction Arterial spin labeling (ASL) provides a method by which to non-invasively measure the spatial and temporal characteristics of local tissue perfusion. The standard methods employ spatial tagging schemes to invert the blood [1]. However, spatial tagging schemes depend upon the spatial separation of the inversion and imaging regions and are thus susceptible to artifacts due to spatial variations in the transit delay (the time between the tagging of the blood and its arrival in a voxel) and low SNR from T1 decay during the transit delay. An alternative approach is Velocity Selective ASL (VS-ASL) [2,3] in which the tagging depends upon velocity profile such that decelerating spins up to a specified cutoff velocity Vc are tagged. This method was introduced because it has the advantages that the tagging is insensitive to transit delays, and selective ranges of velocity can be encoded. However, in this abstract we point out that another distinct advantage of VS-ASL is that it can be velocity encoded in any direction, thereby allowing for the measurement of perfusion at arbitrary angular resolution. This then facilitates the reconstruction of the local perfusion field, which we show can be characterized by a perfusion tensor P, from which can be derived quantities related to the structure of the local perfusion field, such as the mean perfusion, the perfusion anisotropy, and the principal directions of flow feeding each voxel. We demonstrate this new method, Perfusion Tensor Imaging (PTI), in a normal volunteer.

Theory Velocity selectivity was achieved in the manner described by Norris [4] using a $90_x - g_k - 180_y - g_k - 90_x$ pulse train where g_k is a gradient in the k'th direction. For gradients of magnitude G width δ, and separation Δ, this produces a velocity dependent longitudinal magnetization $M_z = M_0 \cos(\beta)$ where $\beta = \gamma G \Delta \delta v_g$ where $v_g = v \bullet g = v\cos(\theta)$ is the projection of the velocity v onto the encoding gradient direction.. The cutoff velocity $v_c \equiv \pi/(\gamma \delta \Delta G)$ [3] then defines a cutoff angle $\theta_c = \cos^{-1}(v_c/v)$ below which spins are tagged. Thus the sensitivity VS-ASL sensitivity to velocity can be phrased in terms of a sensitivity to flow direction, which can be manipulated by changing v_c , and has the shape of a cone, as shown in Figure 1. Tagged spins are represented by the area inside the cone. By using a relatively high v_c and rotating the orientation of the g_k, the angular distribution of the perfusion can be mapped in an analogous fashion to diffusion tensor imaging. The resulting angular measurements are real and positive, and, because velocities both parallel and anti-parallel to the gradient direction are encoded identically, are also symmetric. The perfusion measurements can thus be characterized by a perfusion tensor ***P***, from which can then be derived estimates of the mean perfusion from Trace(P), measures of anisotropy from the eigenvalue decomposition of P, and an estimate of the primarily direction of perfusion to a voxel from the principal eigenvector of P.

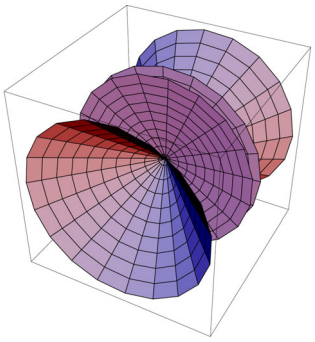
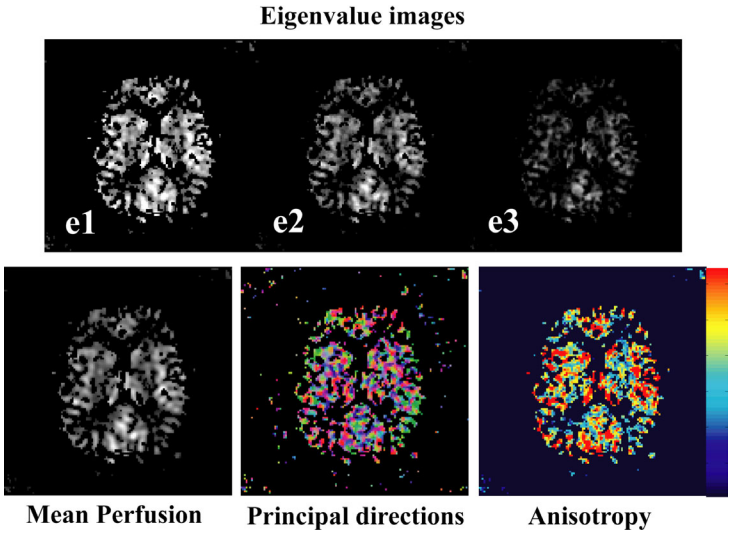


Figure 1. The VS-ASLvelocity directional sensitivity for ratios Vc/V=.1 (outer cone) and Vc/V=1 (inner cone). Outer cone is flat, and thus has essentially no directional sensitivity but inner cone from higher Vc does.

Methods Images were collected on a GE SIGNA Lx Clinical Imager with EchoSpeed gradients, using a standard head coil receiver. The VS-ASL pulse sequence was that proposed by Wong [2] using a spiral acquisition and modified to encode velocity in arbitrary directions, using the following imaging parameters: FOV=24cm, thickness=8mm, matrix=64x64, Vc=17cm/s, TR=2500ms, Tv=1000ms, reps=100. Velocity encoding directions were the 6 non-coplanar directions used in the standard DTI scheme [5]. A complication is that VS-ASL encoding gradients produced eddy currents that distort the images. This effect can be reduced by collecting reference images with short TI's that minimize perfusion effects while retaining the eddy current effects. Reference images were collected with the same parameters as above except for TR=500ms, Tv=50ms and used retrospectively to correct the data.

Results The eigenvalue decomposition of the perfusion tensor P is shown in the top of Figure 2. Voxels containing unphysical negative eigenvalues were set to zero. On the bottom is shown three images derived from the eigenvalues. The mean perfusion (bottom,left) is derived from Tr(***P***), the trace of ***P***, and clearly is representative of a normal perfusion map [eg, 1]. The principal directions of flow (bottom, middle) are determined from the principal eigenvector and color coded according to direction by mapping (A/P,R/L,S/I) onto (r,g,b). This map reveals the principal direction from which each voxel is fed, and thus contains heretofore unavailable information about the local perfusion. The perfusion anisotropy (bottom, right), here expressed as the variance of the eigenvalues, clearly demonstrates spatial variations in anisotropy.

Conclusion We have demonstrated the use of VS-ASL in the formation of perfusion tensor images (PTI) from which can be derived the mean perfusion, the principal directions of inflow, and the inflow anisotropy in each voxel. This has potential import in applications for a variety of applications for which the structure of the local inflow is desired. This technique can be extended to an arbitrary number of encoding directions to perform high angular resolution perfusion (HARP) imaging for the investigation of arbitrarily complex local inflow structure.



References

[1] Wong, et.al. NMR in Biomedicine
[2] Wong, et.al. ISMRM 2002, p621,
[3] Duhamel, et. al. MRM 50:145 (2003)
[4] Norris, et.al. JMR p231, (1999)
[5] Basser, et. al.
[6] Frank, MRM. 47(6):1083-99 (2002)