Perfusion is the term applied to capillary blood flow in tissue. Since the blood carries oxygen and nutrition to the tissue through the capillaries, perfusion is important in maintaining tissue viability. The study of brain perfusion has clinical applications due to the changes in perfusion associated with several neurological diseases. Diagnosis, lesion characterisation and follow-up of treatment in oncology, depression, dementia and acute ischaemic stroke are examples where assessment of perfusion is of value.

A widespread approach of cerebral perfusion measurements using MRI in combination with an exogenous contrast agent is the dynamic susceptibility contrast MRI (DSC-MRI) method [1,2]. By tracking the first passage of the contrast agent/tracer through the blood vessels in the brain using rapid imaging and by applying kinetic models for intravascular tracers [3-4] perfusion parameters such as cerebral blood flow (CBF) in ml/(min 100 g), cerebral blood volume (CBV) in ml/100 g and mean transit time (MTT) in seconds can be calculated (Fig. 1).

This approach relies on the fact that the exogenous paramagnetic contrast agent produces local magnetic field gradients that extend from the vascular compartment into the surrounding tissue, even if the contrast agent is not present in the extravascular space. The contrast agent remains in the vascular compartment and this compartmentalization creates susceptibility gradients. The gradients cause local dephasing of the spins, leading to signal loss in T2*-weighted magnetic resonance (MR) images during the passage of the contrast-agent bolus. Signal loss is also seen in spin echo pulse sequences due to intravascular T2 shortening in combination with spin diffusion in the contrast-agent-induced magnetic field gradients.

Normally it is assumed that the tissue concentration of the contrast agent, C, is proportional to the change in T2* relaxation rate, \( \Delta R_2^* \) [5], i.e.:

\[
\Delta R_2^* = kC
\]
The signal, $S_c$ after contrast-agent administration, for gradient echo pulse sequences, can be expressed as [6]:

$$S_C(t) = S_0 e^{-TE\Delta R_2*}$$  \hspace{1cm} [2]

where $S_0$ is the baseline signal and TE is the echo time. An analogous equation can be written for the relationship between $S_C$ and $\Delta R_2*$ for spin echo sequences.

The combination of Eqs. 1 and 2 gives an expression for the concentration of the contrast agent:

$$C(t)=\frac{1}{kTE} \ln \left( \frac{S_C(t)}{S_0} \right)$$  \hspace{1cm} [3]

In most applications, the proportionality constant $k$ is unknown, and the concentration is given in arbitrary units.

The so-called tissue impulse residue function, $R(t)$ describes the retention of a tracer in the tissue [3,7]. Thus, $R(t)$ is the fraction of the injected tracer still present in the vasculature at time $t$ after an infinitely short injection of tracer into the tissue-feeding artery. The tissue impulse residue function can be expressed as:

$$R(t)=1-\int_0^t h(t)dt$$  \hspace{1cm} [4]

where $h(t)$ is the transit-time distribution and $h(t)dt$ is the fraction of tracer that leaves the capillary system on the venous side during the time interval $t+dt$. At time zero all tracer is still present in the local microvasculature i.e. the tissue residue function has a value 1.

In practice the arterial tracer bolus will not arrive to the tissue as a delta function due to the duration of the injection of the tracer, and the transport of the tracer from the injection site, through the vasculature to the brain. Consequently, the measured concentration-versus-time curve does not reflect the response to an infinitely short arterial bolus. Instead, the concentration curve is the convolution of CBF$\times$R(t), and the concentration-versus-time curve in the tissue-feeding artery, i.e., the arterial input function (AIF) [7-9]. A correction factor given by $kH=1- H_{\text{large}} /[\rho(1\,-\,H_{\text{small}})]$, where $H_{\text{large}}$ and $H_{\text{small}}$ are the haematocrit values in large and small vessels, respectively, and $\rho$ is the density of brain tissue [8], is often introduced. The inclusion of haematocrit values into $kH$ is due to the fact that the tracer is not distributed in the whole-blood but in the plasma volume:

$$kH C(t) = CBF R(t) \otimes AIF(t) = CBF \int_0^t AIF(\tau)R(t-\tau)d\tau$$  \hspace{1cm} [5]

By measuring the arterial input function, the CBF can be determined by deconvolution, as the initial height of the product of CBF and R(t), due to the fact that R(0)=1. Furthermore, the MTT can be determined by Zierler’s area-to-height relationship, according to Eq. 6 [10]:
\[ MTT = \frac{\int_0^\infty R(t)\,dt}{\max[R(t)]} \]  

where \( \max[R(t)] \) is the peak value (theoretically the initial value) of the tissue impulse residue function. Furthermore, the blood volume can be calculated by taking the product of the blood flow and the mean transit time [3,11].

\[ CBV = CBF \cdot MTT \]  

The cerebral blood volume can be derived from Eqs. 5-7:

\[ CBV = \frac{(1 - H_{\text{large}}) \int_0^\infty C(t)\,dt}{\rho (1 - H_{\text{small}}) \int_0^\infty AIF(t)\,dt} \]

Another important concern is that the signal depends not only on the concentration of the tracer but also on the vascular composition and geometry [12]. Fortunately, the main difference in sensitivity appears to be present between artery and tissue, and not between different types of tissue capillary systems. However there exists a non-linearity \( \Delta R2^* \)-versus-concentration in blood making the quantification difficult [13].

REFERENCES