Appendix E1

Concentration-time curves for the artery (arterial input function [AIF]), vein (venous output function [VOF]), and brain tissue were generated by using previously described methods (16,17). An AIF, $C_a(t)$, was generated by using the following gamma-variate function:

$$C_a(t) = \begin{cases} 0 & t \le t_o \\ C_0 (t - t_0)^a e^{-(t-t_o)/b} & t > t_o \end{cases},$$

With several constant values (C_0 of 1.0, t_0 of 12 seconds, a of 3.0, and b of 1.5 seconds). Tissue curves, $C(t)$, were then generated by convolution of the AIF and the residue function, $R(t)$.

$$C(t) = k \cdot \mathrm{CBF} \cdot C_a(t) \otimes R(t),$$

where k is a constant factor of hematocrit difference between large (H_{LV}) and small (H_{SV}) vessels (23).

$$k = \frac{1 - H_{\mathrm{SV}}}{1 - H_{\mathrm{LV}}}.$$

The convolution was performed at a temporal resolution of 0.1 seconds to minimize discretization artifacts. Three types of $R(t)$—exponential, linear, and box shaped—were used to simulate various flow dynamics:

$$\text{Exponential: } R(t) = e^{-\frac{1}{\mathrm{MTT}}},$$

$$\text{Linear: } R(t) = \begin{cases} 1 - \dfrac{t}{2 \cdot \mathrm{MTT}} & t \le 2 \cdot \mathrm{MTT} \\ 0 & t > 2 \cdot \mathrm{MTT} \end{cases},$$

$$\text{Box shaped: } R(t) = \begin{cases} 1 & t \le \mathrm{MTT} \\ 0 & t > \mathrm{MTT} \end{cases}.$$

A VOF, $C_v(t)$, was also generated by convolution of the AIF with an exponential $R(t)$, t_0 of 16 seconds, and MTT of 6.0 seconds.

All the curves were sampled with a sampling time of 2 seconds and time length of 60 and 100 seconds for CT perfusion and perfusion-weighted imaging, respectively. Because the concentration of the iodinated contrast agent in CT perfusion has a linear relationship with Hounsfield units, all the concentration-time curves in CT perfusion, including the AIF, VOF, and tissue, were multiplied by a constant value of 100 to convert into Hounsfield units. A constant value of 50 was used for AIF to simulate the partial volume effect. Baseline values of 40 HU for AIF and VOF and 36 HU for tissue were applied. This scaling and offsetting was conducted to mimic clinical data. For perfusion-weighted imaging, all the concentration-time curves, $C(t)$, were converted into signal-time curves, $S(t)$, by using the following equation:

$$S(t) = S(0) e^{-kC(t)\mathrm{TE}},$$

where $S(0)$ is 200 in arbitrary units and echo time (TE) is 50 milliseconds. A proportionality factor k was set to values described in a previous study (17). For vessel signals (AIF and VOF) and tissue signals, it was determined that k achieved 80% and 40% peak signal decreases, respectively.

In CT perfusion, the phantom data were embedded in a real-brain CT image. Simulated Gaussian noise was added to each pixel to achieve a realistic signal-to-noise ratio of 5.0. This value of the signal-to-noise ratio was determined by region of interest measurement of the cortex in a previously obtained clinical CT perfusion image, scanned at 80 kVp, 200 mA, and 1-second gantry rotation.

In perfusion-weighted imaging, the phantom data were embedded in a real dynamic susceptibility contrast perfusion-weighted image of the brain. Gaussian noises were added to complex data (real and imaginary parts) for all the curves to achieve a signal-to-noise ratio of 40. This value of signal-to-noise ratio was also measured by using the region of interest in a previously obtained 1.5-T gradient-echo echo-planar image in an actual patient, with the following parameters: repetition time msec/TE msec, 1500/50; matrix, 256 × 256; and field of view, 240 mm.

As each tissue section contained 7 × 7 quadratic tiles, with each tile measuring 32 × 32 pixels for CT perfusion and 16 × 16 pixels for perfusion-weighted imaging, each tile contained 1024 (for CT) and 256 (for MR) noise realizations of the same hemodynamic parameters.

Reference

23. Wirestam R, Andersson L, Ostergaard L, et al. Assessment of regional cerebral blood flow by dynamic susceptibility contrast MRI using different deconvolution techniques. Magn Reson Med 2000;43(5):691-700.