Normalised Double Inversion Recovery for Quantification of Cerebral Tissue Proportional Density

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Introduction The double inversion recovery (DIR) method [1], when applied in imaging of the brain, allows images of a single tissue type (grey matter(GM), white matter(WM) or cerebrospinal fluid (CSF)) to be obtained. These images are not quantitatively comparable to each other, as the different T_1 values of the three tissue types lead to differing longitudinal magnetisation available for selective image formation. These differences are compounded by variation in signal intensity due to coil sensitivity profiles. We propose a method that overcomes these difficulties to acquire normalised maps that provide estimates of the proportional density of GM, WM and CSF in the brain. This method is potentially useful for studies of brain morphology such as investigations of grey matter atrophy and of cortical abnormalities associated with conditions such as multiple sclerosis and epilepsy.

Data acquisition DIR images were acquired using a Philips Achieva 3-T scanner, a 3D fast spin echo (FSE) sequence, with FOV 240 mm × 240 mm, 60 transverse slices of thickness 2.1 mm, matrix 128 × 128, TR = 8000 ms, TE = 8.6 ms, TSE factor = 75, profile order: low-high, turbo-direction: radial. Two adiabatic inversion pulses are used, the inversion times, TI_1 and TI_2 used for each tissue-specific acquisition are shown in table 1. Proton density images were acquired by switching off the two inversion pulses, but keeping all other acquisition parameters the same. The total acquisition time for the sequences is 24 minutes.

Tissue-specific image	TI_1 (ms)	TI_2 (ms)
Grey matter	3120	420
White matter	3700	800
CSF	1854	404

Table 1. Inversion times used for tissue-specific images

Data analysis We take the ratios of GM, WM and CSF images to the proton density image in order to remove the effects of sensitivity inhomogeneity due to instrumentation limitations. These ratio images are denoted as $g(\mathbf{r})$, $w(\mathbf{r})$ and $c(\mathbf{r})$ respectively. We then use MRIcro [2] to make a mask of the brain from the grey matter image. We assume that the brain is fully occupied by GM, WM or CSF, the three well-defined tissue classes. Therefore an equation for each voxel in the brain can be formulated as

$$\alpha\, g(\mathbf{r}) + \beta\, w(\mathbf{r}) + \gamma\, c(\mathbf{r}) = 1, \qquad (1)$$

where α, β and γ are the correction factors accounting for longitudinal magnetisation differences for grey matter, white matter and CSF, respectively. As the sensitivity inhomogeneity effect is already taken account of, and with the assumption that T_1 values for each tissue type do not depend on position in the brain, the weightings α, β and γ are global values across the whole brain, not dependent on the voxel's position $\mathbf{r}$. Each voxel therefore provides a different instance of the equation, thereby giving us the means to derive optimal values for each weighting factor. The criterion for choosing optimal values of α, β and γ is that the mode of the distribution $\alpha\, g(\mathbf{r}) + \beta\, w(\mathbf{r}) + \gamma\, c(\mathbf{r})$ for all voxels in the brain is 1. In theory, (1) is an over-determined problem as there are only three unknown parameters but as many as millions of equations. Least-square methods could be used to solve such problems. However, we have found that as the distributions of $g(\mathbf{r})$, $w(\mathbf{r})$ and $c(\mathbf{r})$ are asymmetric, the result from the least-squares method deviates from the optimum criterion mode = 1. Therefore we use a brute force algorithm to look for the optimal values of α β and γ with sum images of $\alpha\, g(\mathbf{r}) + \beta\, w(\mathbf{r}) + \gamma\, c(\mathbf{r})$ generated multiple times with different α, β and γ values using multiple start points.

Results We present 3D DIR images from a volunteer in Fig.1 and normalised proportional density images in Fig.2. Fig.3 is the histogram of the sum images. We used the above method to measure proportional densities of GM, WM and CSF from 5 volunteers (age ranged from 24 to 59) and their results are presented in Fig.4.
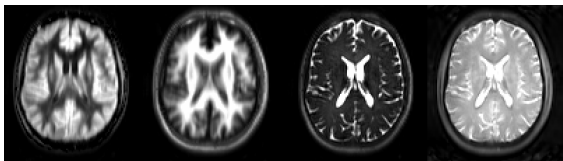

Fig.1. From left to right, images of GM, WM, CSF and PD of a brain acquired using the described DIR method
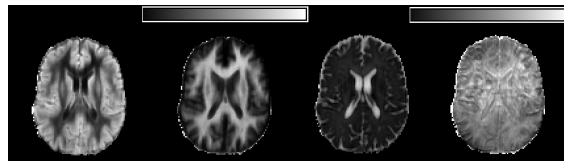

Fig.2. From left to right, proportional density images of GM, WM and CSF displayed on a scale of [0, 1], and their sum (right) on a scale of [0.8, 1.2]
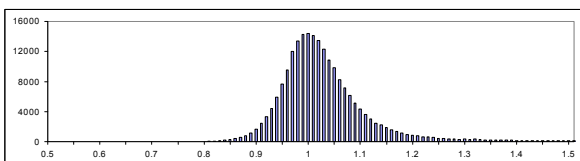

Fig.3. Histogram from the sum of normalised GM, WM and CSF proportional density images, clearly showing that the mode of the distribution is 1.
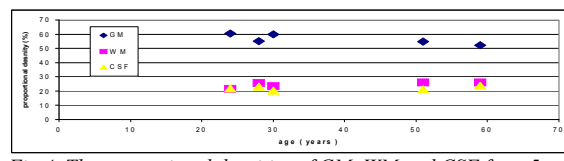

Fig.4. The proportional densities of GM, WM and CSF from 5 volunteers, in which decrease of GM with age can be seen at a rate of -0.184 % per year, with Spearman's rank correlation coefficient $\rho = -0.9$ and $p = 0.0417$.

Discussion The DIR method is based on the assumption that a single T_1 exists in each component of the brain and that inversion is fully efficient; these approximations affect the accuracy of our DIR method to an extent, as indicated by the residual structure in the sum image shown in Fig. 2.. We used a simple brute force algorithm to obtain the optimal values of the correction factors, which works well. More intelligent algorithms could be applicable to improve the efficiency of calculation. Once the correction factors α, β and γ are known the percentage of GM, WM and CSF in each voxel and the whole brain are easily obtainable, which could be useful in studies of tissue or regionally-specific atrophy and in identifying differences between disease groups. The advantage of our method over the conventional approach of brain tissue segmentation is that the separation into tissue type is achieved at the point of imaging, and is therefore guaranteed to deal correctly with partial-volume effects.

References **1**. Redpath TW, Smith FW. Technical note: use of a double inversion recovery pulse sequence to image selectively grey or white brain matter. Br J Radiol 1994; 67:1258-63 **2.** http://www.sph.sc.edu/comd/rorden/mricro.html