Signal to noise ratio of high b-value diffusion weighted images is improved using computed diffusion weighted imaging

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Introduction: There is growing interest in the use of diffusion-weighted images (DWI) for the detection of cancer as tissue dependent properties such as the diffusion coefficient (D), spin-spin relaxation time constant (T2) and fat content combine to provide excellent contrast between diseased and healthy tissue [1]. The DWI signal at each voxel location is described by the well-known Stejskal-Tanner equation, $S(b) = S(0)e^{-bD}$, where b encapsulates the parameters of the diffusion sensitizing gradients. A previous report [2] has demonstrated the use of computed DWI (cDWI) in which multiple b-value images may be used to calculate apparent diffusion coefficient (ADC) maps that are then incorporated into the above equation to compute DW images at any arbitrary b-value. This technique provides increased control over the contrast between diseased and healthy tissue (see figure 1). However, this study did not provide any detail on which b-values should be acquired for ADC calculation, nor did it provide a theoretical model for the noise of cDWI images and its dependence on b-value. The aim of the current report is to elucidate these points using $CuSO_4$ phantom studies.

Theory: Previous reports [3] have demonstrated that for a total of N possible DW image acquisitions of a slice, the optimum acquisition scheme is a two b-value measurement to minimize the error in subsequent ADC calculations. For a tissue with diffusion coefficient D_0, m signal averages should be acquired at b = 0 s mm^{-2} and n signal averages at $b_{opt} = 1.25/D_0$, where m + n = N and n / m ~ 3. This is clearly difficult to achieve over the entire field of view due to the heterogeneity of D_0 although we have found through simulations that using the mean of previously reported values is sufficient. Provided cDWI images are acquired using this regime it is then possible to estimate the noise variance of images calculated at b_c using propagation of errors:

$$\sigma_c^2(b_c) = \sigma_0^2\left(\frac{S(b_{opt})}{s(0)}\right)^{2x}\left\{(1-x)^2 + \frac{1}{3}x^2\left(\frac{S(b_{opt})}{s(0)}\right)^{-2}\right\} \quad (1)$$

σ_0^2 is the variance of noise of the conventionally acquired b = 0 s mm^{-2} image and $x = b_c/b_{opt}$. Expression (1) makes the assumption that noise is Gaussian distributed and its variance is constant across all acquired b-value images. These assumptions have been tested and verified for these phantom studies as shown in figure 2.

Method: The standard deviation of image noise in conventional DWI and cDWI was measured over a range of b-values using $CuSO_4$ phantom studies. A restriction was applied that the acquisition time of both techniques was the same. This reduced the number of signal averages (NSA) that could be acquired for cDWI which requires an extra b = 0 s mm^{-2} measurement for ADC calculation. An optimal b-value for two-point calculation of the phantom ADC used in cDWI, b_{opt}, was derived using the entire range of b-values from the conventional imaging and the methods described in [3]. All other scanning parameters were kept constant: field of view (FOV) = 300 x 300 mm^2, slice thickness (ST) = 5.0 mm, repetition time (TR) = 1100 ms, echo time (TE) = 228 ms and bandwidth in the read direction (BW) = 1812 Hz/px. A plot of the theoretical noise curve (equation 3) was also shown for the calculated b_{opt} and σ_0^2 values.

Results: Using the convention DWI data a mean ADC of 2.05×10^{-3} mm^2 s^{-1} was calculated for the phantom suggesting that the optimum b-value for two-point measurements of ADC is $1.25/2.05\times10^{-3} = 609.76$ s mm^{-2}. The computed DWI was performed using acquired b-values of 0 and 600 s mm^{-2} and the standard deviation of image noise for a range of b-values was measured as demonstrated in figure 2. For the conventional data (black line) noise is roughly constant across all b-values with a mean of 2.46. Conversely, the image noise of the computed images appears to vary across the range of b-values as predicted by equation 3 (dotted curve) and is larger than that for the acquired images when b < 700 s mm^{-2} and smaller when b > 700 s mm^{-2}.

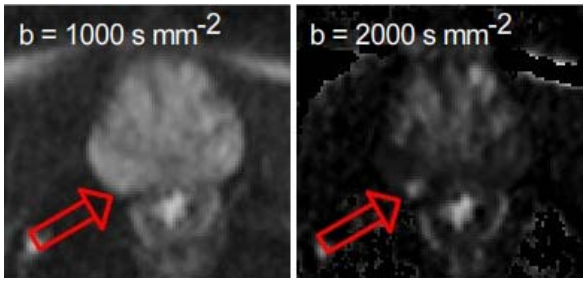

Figure 1. An example demonstrating the value of computed diffusion weighted imaging. Shown on the left is a conventionally acquired b = 1000 s mm^{-2} image of the prostate showing little sign of cancerous involvement in the peripheral zone. On the right is a computed b = 2000 s mm^{-2} where improved contrast has revealed a cancer that was previously undetected.

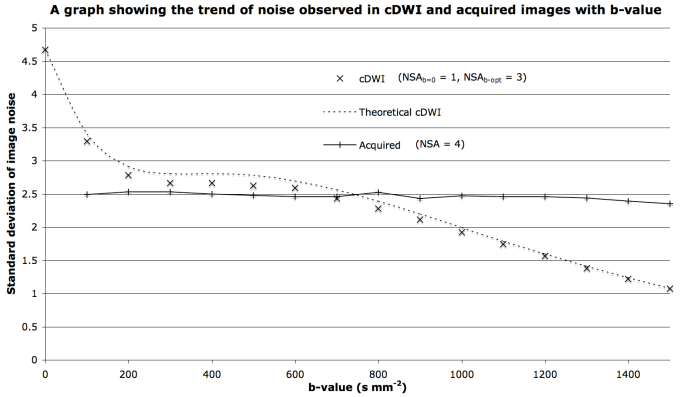

Figure 2. A plot of the standard deviation of noise for both conventionally acquired (+) and computed DWI imaging (x) where the acquisition time is kept the same for both. It is seen that noise remains constant across b-values for the conventionally acquired images whereas for cDWI it reduces with b-value as predicted by equation 1 (dotted curve). It is seen that for high b-values (>700 s mm^{-2}) the SNR of cDWI is improved compared to conventional imaging.

Conclusion: An optimal acquisition scheme for computed diffusion weighting imaging (cDWI) consists of 3 acquisitions at $b_{opt} = 1.25/D_0$ for every single acquisition at b = 0 s mm^{-2} for a tissue with diffusion coefficient D_0. If cDWI data are acquired in this way then it is possible to improve the SNR of high b-value images compared to conventionally acquired DWI, as predicted by equation 1. This is clearly demonstrated in figure 2 where the b-value dependent noise variance of cDWI falls below that of conventional DWI, which remains roughly constant over all b-values, when b is greater than 700 s mm^{-2}. This makes cDWI a potentially powerful tool for obtaining good image contrast for cancer detection at high b-values (figure 1) without having to forfeit the same SNR and noise flooring problems as with conventional DWI (figure 2).

References: [1] Takahara *et al.*, Radiat Med. 22(4):275-82, 2004. [2] Blackledge *et al.*, Proc 17th Annual Meeting ISMRM 2009 (4005). [3] Bito *et al.*, Proc 3rd Annual Meeting ISMRM 1995 (913)

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