Computed Diffusion Weighted Imaging (cDWI) for Improving Imaging Contrast

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Introduction: Higher diffusion weighting (b-values) in diffusion-weighted imaging (DWI) is increasingly used to detect pelvic and bone malignancies. The low apparent diffusion coefficient (ADC) and longer T_2 values of tumours make them visible at high b-values in comparison to the background tissue, which is largely suppressed. However, large b-values (especially > 1000 s mm^{-2}) can cause severe image distortions [1] and the use of longer echo times (TE) reduce the signal-to-noise ratio (SNR) therefore resulting in poorer image resolution [2]. Our aim is to present preliminary results showing the potential advantage in using lower b-values (such as 50, 100 250 and 750 s mm^{-2}) for imaging and then computing higher b-value images (such as b = 1400 s mm^{-2}) using the estimated ADC values with a mono-exponential model:

$$s(b) = s(0)e^{-b \bullet ADC} \quad \text{(eq. 1)}$$

In this study *S(0)* was chosen to be the projected b-0 image based on the ADC calculations made although any other b-value image may be chosen, $s(b_0)$, with the substitution of $b \rightarrow \Delta b$ in equation 1 where $\Delta b = b - b_0$. It should be mentioned here that this model disregards T_2 decay although T_2 dependence could be easily incorporated into the model if the T_2 of the tissues of interest and TE times were known.

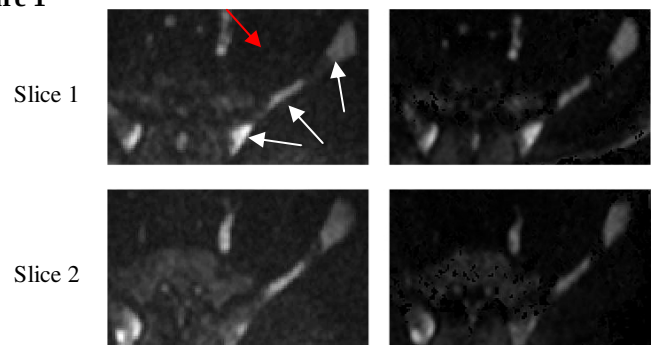
Methods: This approach was applied in two example cases:

Case 1: We compared the SNR of computed and acquired b-1400 images from patient with lymphoma deposits in the iliac bone who was scanned axially using b-values of 50, 100, 250 and 750 s mm^{-2} applied in 3 orthogonal directions for ADC calculations and then again with a single b-value of 1400 s mm^{-2}. Imaging was performed on a 1.5T Siemens Avanto scanner utilizing the double-spin echo technique to minimise eddy current induced distortions [4] with the following acquisition parameters; slice thickness of 5mm, a repetition time (TR) of 3000ms, TE of 80ms and 4 averages. Computed b-1400 images were produced using the in-house software 'DiffusionView' which both calculated the ADC values from the lower b-value range using semi-log plots and incorporated them into the afore mentioned mono-exponential model. The average SNR of the computed and acquired images was computed for 3 foci over 3 contiguous slices by dividing the mean signal within regions of interest (ROIs) around the lesions by the average background noise within an ROI drawn just adjacent to the lesions in a region where no obvious signal was observed (figure 1). One standard deviation in the pixel values within the ROIs was taken to be the uncertainty in measurement.

Case 2: We assessed the viability of extrapolating computed images to a high b-value of 2000 s mm^{-2} in the prostate of a patient with recurrent prostate cancer. ADC values were calculated using b-values of 50, 400 and 800 s mm^{-2} and then a high b-value image was acquired at 1400 s mm^{-2}. In the first case the acquisition parameters were TR = 3100ms, TE = 83ms with 9 averages whilst for high b-value imaging TR = 4211ms, TE = 88ms with 6 averages.

Results: The following are examples of the computed and acquired b = 1400 s mm^{-2} images:

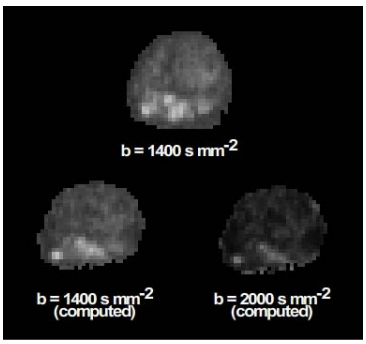
Figure 1



Examples of the acquired (left column) and computed (right column) b = 1400 s mm^{-2} images acquired. There is a noted improvement in the SNR and hence image contrast, between the lymphoma deposits (white arrows) and background tissue (red arrow). The quality of the computed images depends on the accuracy of ADC estimates and many pixels have been zero-filled where the ADC estimates have large uncertainty.

Figure 2

Acquired and computed images of the prostate at different b-values:



Examples in the prostate showing both an image acquired at b = 1400 s mm^{-2} (top) and computed images acquired at b = 1400 and 2000 s mm^{-2}. Equivalent information is observed in the computed images as with the acquired data although they do not suffer from many of the issues associated with high b-value imaging.

The average SNR of the diseased regions in the iliac bone (figure 1) was found to be 5.2 ± 1.7 for computed b-1400 images and 2.0 ± 0.8 for acquired b-1400 images.

Discussion:

Case 1: As observed from both figure 1 and the quantitative analysis, there is a clear improvement in the SNR of the computed images when compared with the acquired b = 1400 s mm^{-2} images which will be beneficial to clinicians for diagnosing sites of disease. However, the success of this technique depends on the accuracy of estimated ADC values and depending on the software used and SNR of acquired images, the model may be ineffective at fitting an ADC value to a particular pixel location. These cases are usually 'zero-filled' (as observed in figure 1) and so there is a potential for misidentification of smaller lesions. We believe that these issues may be overcome by efforts to improve the quality of ADC estimations using various optimisation strategies.

Case 2: In figure 2 it is shown that the computed images provide comparable anatomical information when extrapolated to both b = 1400 and 2000 s mm^{-2} as compared to the acquired image at b = 1400 s mm^{-2}. The computed images did not suffer from the quality issues associated with high b-value imaging such as eddy-current distortion. This technique allows any b-value image to be estimated without re-scanning the patient therefore making it possible to maximise the available contrast between tumour and background tissue based on the ADC measurements [5]. Furthermore, the use of shorter TE enables more averages to be acquired in a similar total scan time.

It is assumed in these examples that the signal attenuation remains mono-exponential although any choice of diffusion model (such as bi-exponential [6]) could be implemented in the image computation assuming the source data supports its use.

Conclusion: Computed diffusion weighted imaging can improve the contrast for detection of pelvic and bone tumours from background tissue at high b-values without the need for acquiring these images where noise and distortions are prevalent. The accuracy of the technique depends on the accuracy of ADC calculations and so we suggest that future efforts should be focused towards optimising ADC estimation techniques rather than optimising image contrast, which may be freely altered using this technique.

References: [1] Haselgrove *et al.*, MRM 36:960 –964, 1996. [2] Jones *et al.*, MRM 42:515-525, 1999. [3] Koh *et al.*, MRMS, 6(4):211-224, 2007. [4] Reese *et al.*, MRM 49:177-182, 2003. [5] Kingsley *et al.*, Concepts Mag. Res., 28A (2):155-179, 2005. [6] Le Bihan *et al.*, Radiol. 161:401-407, 1986

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