A Comparison Between the Spherical Harmonic Transform and the Spherical Spline Approximation for Noise Suppression of DT-MRI Data

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Introduction Spherical harmonic transform (SHT) is used to detect the presence of multiple intra-voxel fibres in diffusion tensor (DT) MRI data with high angular resolution [1,2]. It has also been suggested that, for single fibres SHT may still be beneficial since it can reduce noise by filtering the data across the different DW gradient directions [1]. However, this noise reduction property of SHT has not been demonstrated. Nevertheless, a method of filtering DT-MRI data on the angular domain was recently described based on bicubic spherical splines [3]. This work has investigated the use of SHT as a noise reduction filter of DT-MRI data and compared it with the spherical spline method in simulations and human brain data.

Methods We assume N measured DW signals S_i, corresponding to different DW gradient directions $\boldsymbol{g}_i$ (i=1,2,...,N) and the same b-value b.
SHT The SHT algorithm (http://geoweb.princeton.edu/people/resstaff/simons/software.html) was applied (i) to the apparent diffusion coefficient data $ADC_i = \tilde{}\ln(S_i/S_0)$ /b [1,2] and (ii) to the DW data S_i (i=1,2,...,N). The coefficients of the transform A_{ml} ($|m|\le l$, $0\le l\le L$) were calculated only for even values of l (A_{ml} for odd l were zero by definition due to the antipodal symmetry of the data), and the maximum order L of the transform was such that the total number of coefficients (L^2+L) was maximum without exceeding N. Then the inverse SHT (iSHT) was applied to calculate filtered DW signals S_i' along the directions $\boldsymbol{g}_i$. Different sets of filtered signals were created by varying the maximum order L' of the iSHT. Obviously L' is even and non-zero, to ensure that S_i' is not by definition constant over all $\boldsymbol{g}_i$; thus $2\le L'\le L$.
Bicubic spherical spline This method was used to generate filtered DW signals S_i' along the directions $\boldsymbol{g}_i$, as described in [3].
The filtered signals S_i' (generated by either method) were then used as input to the DT fitting routine. The method was evaluated in simulations and human brain DT-MRI at 1.5T (Infinion, Philips Medical Systems, Cleveland, OH). The DT-MRI scheme used b=1200s/mm^2, 72 uniform DW gradient directions and 6 baseline signals (b=0). Thus maximum order L of the SHT was 8. The simulations used SNR of the baseline signal 25, typical of that in white matter in our human brain DT-MRI datasets. Single-shot DW spin-echo EPI was used.

Results & Discussion

Table 1

	$\lambda_1=\lambda_2=\lambda_3=0.7$		$\lambda_1=1.9$ $\lambda_2=\lambda_3=0.1$	
original	0.053 ± 0.017		0.943 ± 0.009	
spline	0.007 ± 0.018		0.932 ± 0.010	
L'	ADC_i	S_i	ADC_i	S_i
2	0.053±0.017	0.053±0.017	0.944±0.012	0.908±0.010
4	0.053±0.017	0.053±0.017	0.944±0.010	0.942±0.009
6	0.053±0.017	0.053±0.017	0.944±0.009	0.943±0.009
8	0.053±0.017	0.053±0.017	0.944±0.009	0.943±0.009

Table 1 shows simulation results for two types of fibres (noise-free principal diffusivities are denoted as λ_i, i=1,2,3 in 10^{-3}mm^2/s. For each type, mean value ± standard deviation are shown for fractional anisotropy (FA). The columns ADC_i and S_i correspond to SHT performed on the ADC_i or the DW data, respectively. Results using the original data, the spherical splines and iSHT for different orders L' are shown. When SHT is applied to ADC_i, it has no effect on anisotropic data for any $L' \ge 2$; this is in agreement with [1], which shows that in the absence of noise and for a single fibre, SHT on ADC_i and L'=2 is sufficient to estimate anisotropy fully. When SHT is applied to the DW data, it requires $L' \ge 4$ in order not to underestimate strong anisotropy (eg λ_1=1.9, $\lambda_2=\lambda_3$=0.1). Predictably, the original data of the isotropic case ($\lambda_1=\lambda_2=\lambda_3$=0.7) shows noise-induced elevated anisotropy which is not reduced by the SHT methods. Thus, application of SHT on either ADC_i or the DW data, does not reduce noise. In marked contrast, the spherical spline method drastically reduces noise-induced bias at the low anisotropy end, and does not affect estimation of true anisotropy. The difference between SHT and the spherical spline approximation can be explained by the fact that the latter performs piecewise SHT, among automatically chosen segments. The continuity of the splines and their derivatives across neighbouring segments results in data smoothing [4].

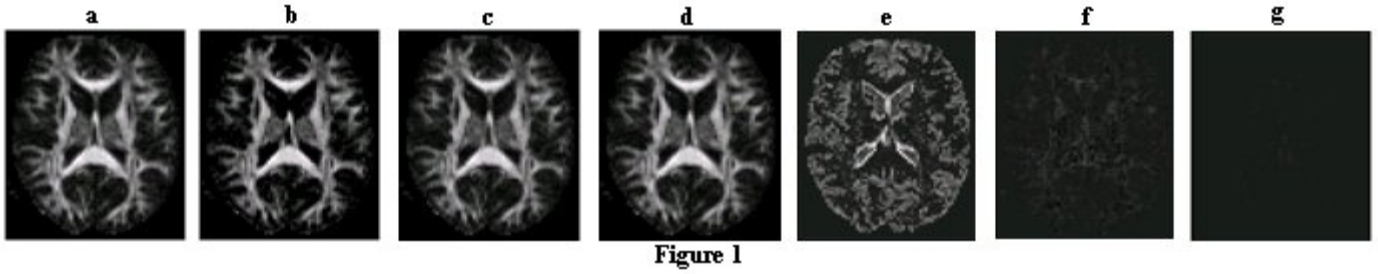

Figure 1

These observations are confirmed by Figure 1, which shows brain FA maps (scaled 0-1). The maps correspond to: (a) the original data, (b) the data preprocessed by the spherical splines, (c) the data preprocessed by SHT of ADC_i with L'=2 and (d) the data preprocessed by SHT of the DW data with L'=4. Difference maps (a)−(b), (a)−(c), (a)−(d) are shown on (e), (f) and (g), respectively, scaled -0.04 to 0.29. While the maps of the SHT-preprocessed data are virtually identical to the map of the original data, preprocessing with spherical splines suppresses anisotropy at low anisotropy areas (within and around sulci and ventricles), improving contrast between low and high anisotropy.

Conclusions SHT [1,2] has been applied as a preprocessing step of DT-MRI data and has been compared to the spherical spline filtering [3]. In addition to its application on the ADCs [1,2], SHT was also applied directly to the DW data. Although both SHT applications created indistinguishable results, SHT of the DW data required a 4-order transform while a 2-order transform was sufficient for SHT of the ADCs for a single fibre. Despite earlier predictions [1], SHT did not perform data denoising. Preprocessing with the spherical splines suppressed markedly noise induced anisotropy without reducing image resolution.

References [1] Frank LR, *Magn Reson Med* 47: 1083-1099 2002 [2] Alexander DC et al, *Magn Reson Med* 48: 331-340 2002 [3] Papadakis NG et al, *Proc 11th ISMRM* 1225 2004 [4] Dierckx P *Curve & Surface Fitting with Splines*, Oxford 1993

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