SINGLE–SHELL DIFFUSION MRI NODDI WITH IN VIVO CERVICAL CORD DATA

Francesco Grussu[1], Torben Schneider[1], Hui Zhang[2], Daniel C. Alexander[2], and Claudia A. M. Wheeler-Kingshott[1]

[1]*NMR Research Unit, Department of Neuroinflammation, Queen Square MS Centre, UCL Institute of Neurology, London, England, United Kingdom,* [2]*Department of Computer Science and Centre for Medical Image Computing, University College London, London, England, United Kingdom*

TARGET AUDIENCE Researchers interested in spinal cord diffusion MRI.

PURPOSE To compare single–shell estimates of Neurite Orientation Dispersion and Density Imaging (NODDI)[1] metrics with double–shell ground truth in the healthy cervical cord *in vivo*.

INTRODUCTION NODDI, a recent model–based diffusion–weighted (DW) MRI technique, distinguishes signals from three compartments (isotropic, intra and extra–neurite) using a two–shell acquisition. To date, it is not clear whether NODDI may utilize standard single–shell diffusion encoding in areas such as the spinal cord, e.g. in retrospective analysis of single *b*–value DW data. This work compares single–shell estimates of NODDI indices, obtained from a reduced model without the isotropic compartment, with optimal double–shell ground truth in the healthy cervical spinal cord.

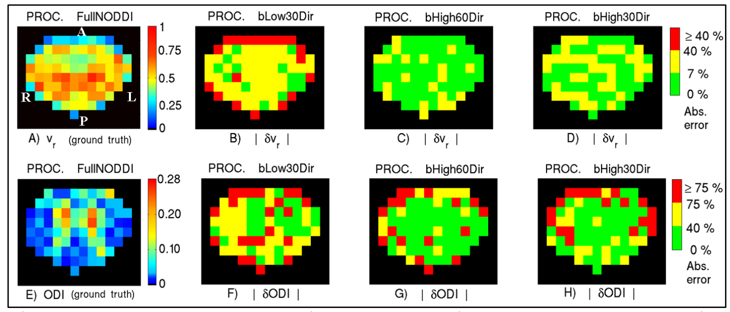

Figure 1: two–shell NODDI metrics (A, E) and single–shell percentage relative errors (in absolute value, B to D, F to H) within the *fitting mask* of 4th slice of S3.

METHODS *Data* We scanned 5 healthy volunteers (subjects S1 to S5, 2 males, mean age 36 years, range 25–47) axially at cervical level (C1–C5) on a clinical 3T Philips Achieva scanner, using a 16–channel neurovascular receiver coil. We followed the NODDI protocol[1] acquiring 6 b = 0 images, 30 DW at b = 711 s mm^{-2} and 60 DW at b = 2855 s mm^{-2}. A cardiac–gated PGSE ZOOM–EPI sequence was employed with TE = 65.5 ms, Δ = 32.2 ms and δ = 20.5 ms for both shells. Other parameters were: TR = 12 RRs, reduced FOV of 64×48 mm^2 for 12 slices, SENSE factor of 1.5, resolution of 1×1×5 mm^3, acquisition time of approximately 35 minutes.

Preprocessing In plane motion was corrected with FSL FLIRT[2]. A cord mask was obtained on the mean b = 0 volume with an active surface method[3] (*fitting mask*) and then cropped to the 6 central slices and eroded (*analysis mask*) to exclude areas with CSF partial volume effects.

Model fitting Four fitting procedures were implemented with the NODDI MATLAB toolbox and run within the *fitting masks* of all subjects. i) *Procedure FullNODDI*: full model fitted to the whole double–shell data set. ii) *Procedure bLow30Dir*: reduced model fitted to the 30 measurements at b = 711 s mm^{-2}. iii) *Procedure bHigh60Dir*: reduced model fitted to the 60 measurements at b = 2855 s mm^{-2}. iv) *Procedure bHigh30Dir*: reduced model fitted to the 30 measurements at b = 2855 s mm^{-2} corresponding to gradient directions of *bHigh60Dir* with minimum electrostatic energy[4], extracted with Camino[5]. The reduced NODDI model omitted the isotropic compartment and was always fitted to single–shell data (procedures *bLow30Dir*, *bHigh60Dir* and *bHigh30Dir*).

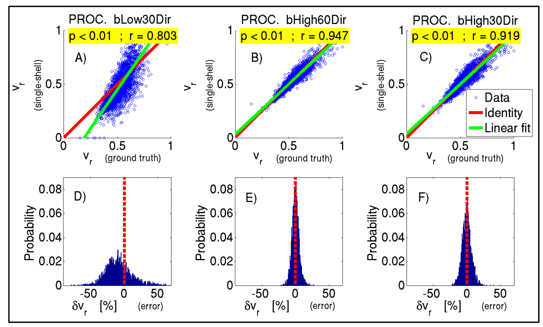

Figure 2: from A to C, scatter plots and correlations in yellow relating double and single–shell values of v_r. From D to F, corresponding distributions of errors.

Analysis Analysis focused on the volume fraction of the intra–neurite compartment (v_r) and the orientation dispersion index (ODI)[1], and aimed to characterize the errors of single–shell metrics with respect to the double–shell ground truth. Firstly, percentage relative errors for single–shell procedures (quantities δv_r and δODI) were calculated. Secondly, voxels within the *analysis masks* of all subjects were extracted. Then, Pearson's correlation coefficients between double–shell values of v_r and ODI and their single–shell estimates were calculated over the extracted voxels. Lastly, the probability distributions of errors P(δv_r) and P(δODI) over the extracted voxels were estimated by calculating the normalized histograms of δv_r and δODI. In practice, we recall that v_r is written as $v_r = (1 - v_{iso})\, v_{in}$, with v_{iso} being the volume fraction of the voxel occupied by isotropic tissue and v_{in} the volume fraction of the non–isotropic tissue occupied by neurites[1]. For single–shell procedures, v_{iso} is constrained to 0 and $v_{in} \equiv v_r$ yields part of the information conveyed by the factor $(1 - v_{iso})$ in procedure *FullNODDI*. For this reason, we adopted metric v_r of ground truth *FullNODDI*, rather than v_{in}, as overall reference.

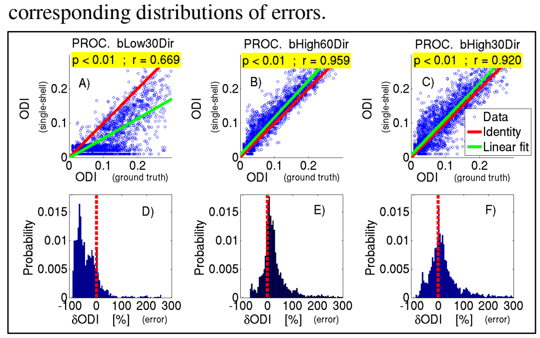

Figure 3: from A to C, scatter plots and correlations in yellow relating double and single–shell values of ODI. From D to F, corresponding distributions of errors.

RESULTS *Metrics* Examples of metrics and errors, consistent with the trend observed in all subjects, are shown in figure 1. Errors were high at the cord border, where CSF partial volume is likely, and stronger for procedure *bLow30Dir* compared to procedures *bHigh60Dir* and *bHigh30Dir*.

Correlations and distributions of errors Figures 2 and 3 show a good agreement between ground truth and single–shell estimates of both metrics, increasing with higher *b*–value and number of measurements (lowest correlations for *bLow30Dir*). Figure 3.A also reveals that low ODI values ranging in [0; 0.18] in the ground truth procedure *FullNODDI* exhibited relatively strong negative errors in procedure *bLow30Dir*, as they were mapped onto a narrower range. Visual inspection proved that those values corresponded to voxels which were mainly in white matter areas. Finally, the distributions of errors showed that both metrics were underestimated in procedure *bLow30Dir*, whereas in procedures *bHigh60Dir* and *bHigh30Dir* only a slight upward bias of ODI was seen.

DISCUSSION In general, fitting of the reduced model to single–shell data could replicate with sufficient accuracy two–shell metrics, provided that the *b*–value is not low. At low *b*–value, we observed stronger biases of metrics and reduced sensitivity of ODI in coherent neural tissue in particular. Furthermore, increasing the number of measurements positively affected single–shell metrics, but not as much as the choice of the *b*–value. As final note, we remark that the absence of the isotropic compartment may in fact limit the specificity of NODDI indices in regions with significant CSF contamination.

CONCLUSION We demonstrated the feasibility of fitting the reduced NODDI model to single–shell data of the cervical spine at high *b*–values. Our analysis also highlighted the extra care needed to interpret single–shell NODDI metrics obtained when *b* is low, since such metrics i) may be biased and ii) may lack sensitivity and/or specificity.

ACKNOWLEDGEMENTS The MS Society in UK, NIHR UCL–UCLH BRC, UCL GCS, all volunteers.

REFERENCES **1.** Zhang H et al. NeuroImage (NI) (2012); 61(4):1000–16. **2.** Jenkinson M et al. NI (2002); 17(2): 825–41. **3.** Horsfield MA et al. NI (2010); 50(2): 446–55. **4.** Jansons KM et al. Inverse Probl (2003); 19(5): 1031–46. **5.** Cook PA et al. Proc ISMRM (2006); p.2759.