Non-uniformity normalization using 3D Canny edges and Legendre polynomial approximation of the bias field: validation on 7T T1W brain images

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MR signal intensity, especially at high field strength, is affected by inhomogeneity, or shading artefacts, manifested as a smooth spatially varying signal intensity distortions. Nonuniformity is attributed to inhomogeneous RF fields, inhomogeneous reception sensitivity and electromagnetic interaction with the object being scanned [1]. Correction of this effect is the key to successful implementation of all MR image analyses, including segmentation, registration and functional modeling of dynamic data. We have developed a new method BiCal (Bias Calculation) for non-uniformity correction that uses two very general assumptions: **a)** bias field is multiplicative (algorithm can be trivially modified for the additive field) **b)** local signal intensity variation within a homogeneous tissue are caused by the bias field.

Methods

We begin by detecting 3D edge surfaces (***3D Canny*** in this implementation). Voxels identified as edges and adjacent voxels are marked. Also marked are background (air) voxels which are identified by signal < 2 times the level of estimated overall image noise. Unmarked voxels constitute a single binary mask **H** that represents all the regions of homogeneous tissue. We next search for a slowly varying analytic scalar field **L(x,y,z) = ln(Bias)** represented as the linear combination of ***3D Legendre polynomials*** up to degree **N**. The value of **N** is defined by the user and controls the maximal spatial frequency of the resulting field. Useful results were achieved for **N** in the range [3-20] for our 7T brain data. Next the logarithm volume $\mathbf{P(x,y,z)=ln(SI_{measured})}$ is calculated. From **a)** $\Rightarrow$ $\mathbf{P=ln(SI_{true})+L}$ and **b)** $\Rightarrow$ $\boldsymbol{dP/dr=dL/dr}$ within **H**. We optimize polynomial coefficients of **L** over **H** so that the analytical partial derivatives {***d/dx, d/dy, d/dz***} provide the least squares fit of discrete partial derivatives of **P.** Discrete derivatives are pre-conditioned by a constrained gaussian smoothing of **P** over **H.** The goal is to remove noise and partial volume effects still remaining after the exclusion of edges. Since the cost function is linear, the corresponding linear system can solved exactly by least squares fitting using SVD. Finally, the resulting field **B(x,y,z) = exp(L)** is normalized so that its average value over non-air voxels is unity**.**

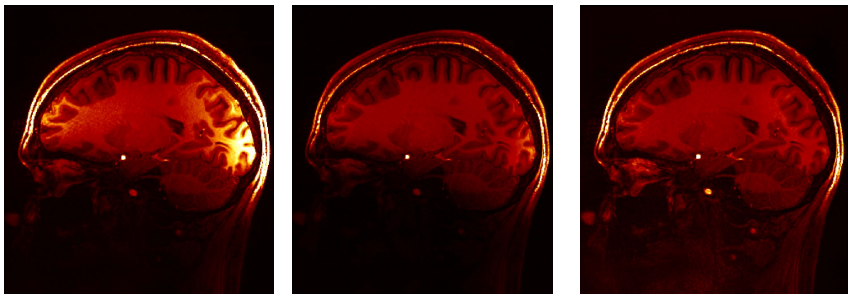
All acquisitions were performed on a 7.0-T imager (Magnetom; Siemens, Erlangen, Germany) by using a volume-transmit 24-element receive coil array (Nova Medical, Boston, Mass). A 3D automatic shimming was first performed by adjusting all fist and second order shim currents, then shim performance was verified by using the interface on the imager console. This was followed by either sagittal or axial 3D T1-weighted magnetization-prepared rapid acquisition gradient-echo (MPRAGE) sequence with repetition time 2.6 msec, echo time 2600 msec, inversion time 1100 msec, 6° flip angle, 248 0.6 mm thick sections, 346 × 323 matrix, 213 × 207-mm field of view, and acceleration factor of two.

	Original	N3	BiCal
v1-WM	0.285	0.173	0.124
v1-GM	0.335	0.225	0.164
v2-WM	0.321	0.211	0.135
v2-GM	0.347	0.238	0.153
v3-WM	0.259	0.195	0.142
v3-GM	0.282	0.226	0.165
Average	0.305	0.211	0.147

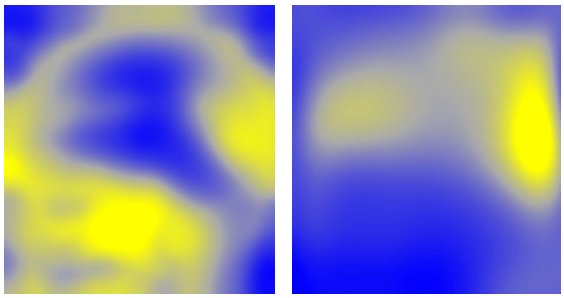
To measure accuracy, 100 closely adjacent pairs of seeds of white and gray matter where placed on 10 equidistant slices of each volume. To avoid trivial solutions, we set the parameters of BiCal and N3 to preserve, within 2%, local tissue contrast $\boldsymbol{C}=SWM_i/SGM_i$, where SWM_i, SGM_i denote the signal intensity of $\mathbf{i_{th}}$ white and gray matter seed. For all cases, all samples and all conditions ***C*** was found to be 1.38+/-0.11. Non-uniformity ***U*** was expressed as StdDev()/Avg() averaged over all seeds. BiCal results were compared to a de-facto standard N3 algorithm [3] with optimized parameters (noise=0.01, fwhm=0.2 and lowest smoothing that minimized ***U*** while keeping ***C*** constraint. In BiCal tests polynomials of degree 12 were used. All software was implemented using MSVC++ 2012 compiler and parallelized using Intel Threading Building Blocks 4.0. Tests were performed on Intel i7-3820QM mobile 2.7GHz processor (4 cores\8 threads 100% engaged). BiCal run took 30 seconds for the optimization part (plus variable time for 3D edge calculation, which is 10 seconds for ***3D Canny***).

Results and discussion

Nonuniformity results ***U*** are presented in the table above. In comparison to N3, BiCal calculated bias field (figure below) far less resembles the underlying anatomy with less chances for producing artefacts, an issue that is of great concern of researchers evaluating MRI non-uniformity algorithms.



Representative result: Original volume, N3-corrected volume, and BiCal-corrected volume



N3 (left) and BiCal bias fields

While some components of the BiCal were proposed in the literature [2], suboptimal methods were used to exclude high-gradient voxels. Inefficient gradient conditioning schemes were also used. Also Legendre polynomials of degree >3 were dismissed as computationally inefficient. Our experiments indicate that for the realistic field of view and image resolution values, Legendre basis of degrees >6 is required. Being fully parallel BiCal implementation would 100% scale on all **multi-core** and **many-core** (f.e Xeon Phi) modern processors. Overall, BiCal method for nonuniformity correction shows promising results, worthy of further investigation.

References:

1. Boyes et al. Intensity non-uniformity correction using N3 on 3-T scanners with multichannel phased array coils. NeuroImage 39 (2008) 1752–1762
2. Styner. 2000. Parametric estimate of intensity inhomogeneities applied to MRI. IEEE TRANSACTIONS ON MEDICAL IMAGING, VOL. 19, NO. 3, P153-165
3. Sled, J.G. et al. 1998. A nonparametric method forautomatic correction of intensity nonuniformity in MRI data. IEEE Trans. Med. Imag. 17, 87–97.