Active contours-driven registration method for the structure-informed segmentation of diffusion MR images

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Abstract

Current applications of whole-brain tractography to diffusion MRI data require highly precise delineations of anatomical structures, which are usually projected from T1 weighted images by registration. In this study, we propose regseg, which is a simultaneous segmentation and registration method that uses active contours without edges extracted from structural images. The contours evolve through a free-form deformation field supported by the B-spline basis to optimally map the contours onto the data in the target space. We tested the functionality of regseg using four digital phantoms warped with known and randomly generated deformations, where subvoxel accuracy was achieved. We then applied regseg to a registration/segmentation task using 16 real diffusion MRI datasets from the Human Connectome Project, which were warped by realistic and nonlinear distortions that are typically present in these data. We computed the misregistration error of the contours estimated by regseg with respect to their theoretical location using the ground truth, thereby obtaining a 95% CI of 0.56–0.66 mm distance between corresponding mesh vertices, which was below the 1.25 mm isotropic resolution of the images. We also compared the performance of our proposed method with a widely used registration tool, which showed that regseg outperformed this method in our settings.

Keywords: active contours, cortical parcellation, diffusion MRI, nonlinear registration, segmentation, susceptibility distortion.

1. Introduction

The accurate delineation of white matter (WM) in diffusion MRI (dMRI) and the fusion of prior anatomical information extracted from a T1 weighted (T1w) image for the same subject are crucial in a range of applications based on tractography, such as the extraction of structural connectivity (Craddock et al., 2013) or tract-based spatial statistics (Smith et al., 2006). However, the precise segmentation of dMRI data and the coregistration of anatomical images are difficult due to several limitations. First, dMRI images have a resolution that is much higher than that of the imaged microstructural features (Basser and Pierpaoli, 1996). Therefore, voxels located in structural discontinuities are affected by partial voluming of the signal sources. Second, dMRI schemes probe the diffusion process within the brain at many angles to obtain diffusion weighted images (DWIs), which are completed by one or more baseline ($b_0$) volumes without

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directional gradients. The extremely low signal-to-noise ratio (SNR) and the high dimensionality of DWIs prevent their direct use in segmentation. The low contrast between grey matter (GM) and WM in the b0 volumes also makes them unsuitable for segmentation. Finally, dMRI images are acquired using echo-planar imaging (EPI) to speed up the acquisition process but at the cost of introducing geometrical distortion, as well as signal degradation and destruction (Jezeard and Balaban, 1995). The artifacts generated have major impacts on the anatomy extracted from dMRI, particularly in certain fiber bundles (Irfanoglu et al., 2012). These limitations prevent segmentation in the native dMRI space, including registration-based approaches using anatomical images as the sources of structural information.

Early attempts to delineate the WM involved thresholding the fractional anisotropy (FA)\(^1\) map. However, the mask and subsequent analyses are highly dependent on the threshold set (Taoka et al., 2009).

To overcome the unreliability of FA thresholding, Zhukov et al. (2003) proposed active contours with edges represented by level sets, which evolve on directionally invariant scalar maps. Rousson et al. (2004) successfully segmented the corpus callosum with region-based level sets based on the eigenvalues of tensors fitted using the dMRI data. Many studies have addressed the definition of appropriate features for clustering, such as the 5D representation proposed by Jonasson (2005). Other efforts include mixed models based on sets of directionally invariant maps (Liu et al., 2007), iterative (Hadjicoplos et al., 2005) and hierarchical (Lu et al., 2008) clustering, graph-cuts (Han et al., 2009), and volume fraction modeling (Kumazawa et al., 2013).

To address the segmentation problem by registration, Saad et al. (2009) used the Pearson’s correlation coefficient to obtain a linear alignment of the T1w and the b0. Similarly, a linear registration method was employed by bregister (Greve and Fischl, 2009), which uses active contours with edges to search for intensity boundaries in the b0 image. The active contours are initialized using surfaces extracted from the T1w using FreeSurfer (Fischl, 2012): the pial surface (exterior of the cortical GM) and the white surface (the WM/GM interface). The b0 image only includes a detectable frontier for the pial surface, so bregister is limited to aligning the cortical layer in this application. This tool has become the standard method because of its proven robustness, although the distortions found in dMRI are nonlinear. To overcome this issue, bregister excludes the regions that are typically warped by artifacts from the boundary search. However, because the distortion is not considered, it must be addressed separately. Nonlinear registration has been performed successfully between T2 weighted (T2w) and b0 images based on their similarity as a unique method for correcting distortion (Kybic et al., 2000; Studholme et al., 2000; Wu et al., 2008; Tao et al., 2009). However, further registration of the T1w and T2w images is still required to map the anatomical information (and for WM segmentation) into the dMRI space.

Therefore, given these issues, a method that simultaneously performs segmentation in the native dMRI space and registration of the corresponding T1w image onto the dMRI data could provide the optimal solution. To the best of our knowledge, this strategy has never been proposed for the application described above. Previously, joint segmentation and registration have been applied successfully to other problems such as longitudinal object tracking (Paragios, 2003) and atlas-based segmentation (Gorthi et al., 2011). The most thorough approach integrates active contours during image registration methods. Unal and Slabaugh (2005), and later Wang et al. (2006), improved an existing method (Yezzi et al., 2003) based on linear registration to the nonlinear case by implementing a free-form deformation field. Droske et al. (2009) reviewed the existing techniques and proposed two different approaches for applying the Mumford-Shah functional (Mumford and Shah, 1989) during simultaneous registration and segmentation by propagating the deformation field from the contours onto the whole image definition. Recently, Le Guyader and Vese (2011) proposed a simultaneous segmentation and registration method in 2D using level sets and a nonlinear elasticity smoother on the displacement vector field, which preserves the topology even with very high deformation. Finally, Gorthi et al. (2011) extended the existing methodologies using a multiphase level set function to register several active contours during the application of atlas-based segmentation.

The hypothesis tested in the present study is that registration and segmentation problems in dMRI can be solved simultaneously, thereby increasing the geometrical accuracy of the process. Thus, we propose

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\(^{1}\)the FA is a scalar parameter of diffusion derived from the dMRI data.
a tool called `regseg` that exploits the prior information of shapes extracted using T1w images to register the anatomical reference to dMRI space, thereby implicitly segmenting these data. The proposed approach uses active contours without edges (Chan and Vese, 2001), which evolve to drive a free-form deformation field of B-spline basis functions. Optimization is performed using a descent strategy with explicit shape gradients (Jehan-Besson et al., 2003; Herbulot et al., 2006). Therefore, unlike most previous methods, `regseg` does not implement level sets. The nonlinear distortion is aligned with one of the imaging axes (see subsection 2.5), so `regseg` includes an anisotropic regularizer for the displacements field proposed by Nagel and Enkelmann (1986). Finally, we evaluated `regseg` using an extension of our instrumentation framework (Esteban et al., 2014), which simulates known and realistic distortions based on dMRI data. The evaluation included a comparison with T2w-registration based (T2B) correction by integrating it within the framework of an in-house implementation of the method.

2. Methods

Let \( \Gamma_R = \{ \Gamma_m : m \in \mathbb{N}, m \leq N_S \} \) be the set of \( N_S \) surfaces extracted from the undistorted T1w image (the reference space \( R \)). We reformulate the segmentation of the distorted dMRI images (the moving space \( M \)) as a registration problem where we search for an underlying deformation field \( U \) such that the structures in \( R \) defined by \( \Gamma_R \) align optimally with their corresponding structures in \( M \):

\[
U : R \subset \mathbb{R}^n \rightarrow M \subset \mathbb{R}^n
\quad r \mapsto r' = r + u(r),
\]

where \( r \) denotes a position in \( R \), \( r' \) is its corresponding location in \( M \), and \( n \) denotes the dimensionality of images. Finally, \( u = u(r) \) is the displacement of every point with respect to the reference domain.

2.1. Cost-function derivation

In a Bayesian framework for registration (Wyatt and Noble, 2003; Pohl et al., 2006; Gass et al., 2014), the mappings \( U \) in (1) are evaluated based on their posterior probability given the observed data \( M \). Let \( \Omega = \{ \Omega_l : l \in \mathbb{N}, l \leq N_L \} \) be the set of \( N_L \) competing regions in which \( M \) is partitioned by the projection of \( \Gamma_R \). Using Bayes’ rule, the posterior likelihood is computed as:

\[
P(U \mid M, \Omega) = \frac{P(M \mid U, \Omega) P(U)}{P(M)}.
\]

where \( P(M \mid U, \Omega) \) is the data likelihood. Because \( \Omega \) are mapped by \( U \), then we simplify \( P(U, \Omega) = P(U) \Rightarrow P(M \mid U, \Omega) = P(M \mid U) \). The best estimate \( \hat{U} \) then satisfies the maximum a posteriori criterion (Bishop, 2006) and it aligns \( \Gamma_R \) into \( M \). First, we assume independence between pixels, and thus we break down the global data likelihood into a product of pixel-wise conditional probabilities:

\[
P(M \mid U) = \prod_l \prod_i P(f_i' \mid U),
\]

where \( f_i' = M(r_i') \) is the feature vector at the displaced position \( r_i' \) (1) in the moving image. For convenience and because it has been shown to be an appropriate approximation (Van Leemput et al., 1999; Cuadra et al., 2005), we introduce two assumptions for each region \( \Omega_i \): 1) the features are i.i.d.; and 2) they can be modeled by multivariate normal distributions with the parameters \( \{ \mu_i, \Sigma_i \} \) for each region \( \Omega_i \):

\[
P(M \mid U) = \prod_l \prod_{i \in \Omega_i} \frac{1}{\sqrt{(2\pi)^D |\Sigma_i|}} e^{-\frac{1}{2} (f_i' - \mu_i)\Sigma_i^{-1}(f_i')}.
\]
using $D^2_l(f)$ to denote the squared Mahalanobis distance of $f$ with respect to the descriptors of region $l$ as $D^2_l(f) = (f - \mu_l)^T \Sigma_l^{-1} (f - \mu_l)$. $C$ is the number of channels comprised in the image $M$. In fact, the projection of $\Gamma_R$ onto $M$ is an implicit segmentation model, for which the covariance matrix $\Sigma_l$ of each region is minimized.

The smoothness of the resulting displacements field is induced by a Thikonov regularization prior:

$$P(U) = \prod_r p(u) = \prod_r p_0(u) p_1(u),$$

with

$$p_0(u) = \mathcal{N}(u | 0, A^{-1}),$$

$$p_1(u) = \mathcal{N}( \nabla u | 0, B^{-1}),$$

which requires that the distortion and its gradient have zero mean and variance governed by the matrices $A$ and $B$. Finally, the maximum a posteriori problem is adapted to a variational problem where we search for the minimum of an energy functional by applying $E(u) = -\log \{ P(M \mid U) P(U) \}$:

$$E(u) = -\log \prod_l \mathcal{N}(f' \mid \mu_l, \Sigma_l) p_0(u) p_1(u) =$$

$$= \text{Const.} + \sum_l \int_{\Omega_l} D^2_l(f') \, dr + \int_{\Omega} \frac{1}{2} [u^T A u + (\nabla u)^T B (\nabla u)] \, dr.$$  

This expression is the dual of the Mumford-Shah functional that corresponds to the framework of active contours without edges (Chan and Vese, 2001) with the anisotropic regularization term of Nagel and Enkelmann (1986).

### 2.2. Numerical Implementation

#### Deformation model. Since the vertices of the surfaces $\{ v_i : v_i \subset \Gamma \}_{i=1}^{N_v}$ are probably located off-grid, it is necessary to derive $u_i = u(v_i)$ from a discrete set of parameters $\{ u_k \}_{k=1}^{K}$. Densification is achieved using a set of associated basis functions $\psi_k$ (8). In our implementation, $\psi_k$ is selected as a tensor-product B-spline kernel of degree three.

$v'_i = v_i + u_i = v_i + \sum_k \psi_k(r) u_k.$

#### Optimization. To find the minimum of the energy functional (7), we propose a gradient-descent approach with respect to the underlying deformation field using the following partial differential equation (PDE):

$$\frac{\partial u(r,t)}{\partial t} = -\frac{\partial E(u)}{\partial u_k},$$

where $t$ is an artificial time parameter of the contour evolution and $u_k$ are the parameters that support the estimate $\hat{U}$ of the transformation at the current time point. Let us assume that the anisotropy is aligned with the imaging axes to simplify (7) as expression (A.1) in Appendix 1, and thus to compute its derivative (9):

$$\frac{\partial E(u)}{\partial u_k} = \frac{\partial}{\partial u_k} \left\{ \sum_l \int_{\Omega_l} D^2_l(f') \, dr + \int_{\Omega} \frac{1}{2} [\alpha \cdot u^{\circ 2} + \beta \cdot (\nabla u)^{\circ 2}] \, dr \right\},$$

where $u^{\circ 2} = u^T \cdot u$. Then, the data and regularization terms are split and discretized to compute their derivatives. The computation of the explicit shape gradients at each $v'_i$ is illustrated in Figure 1. Then, (10) can be reformulated as (see Supplemental Materials, equations SM6-SM12):
Figure 1: The active contours are defined as the interfacing surfaces of the competing ROIs $\Omega_i$, which are represented in green and dark blue in this close-up. They evolve iteratively following their inward normals $\hat{n}_i$ at each vertex $v_i$ of the mesh. The gradient speeds $\bar{s}_i$ drive registration, which are computed as the disparity of the data energies with respect to the two limiting regions of $M(v_i)$, the features of the image $M$ in the location of vertex $v_i$ (see Supplemental Materials, equation SM5). In this figure, $\bar{s}_1$ is written in the lower box, with $\Omega_{wm}$ in the inner limiting region, $\Omega_{gm}$ the outer region, and $w_{0,1}$ is the relative area associated with vertex $v_1$ with respect to the total area of surface $\Gamma_0$. 

$$\bar{s}_1 = w_{0,1} \left[ D^2_{wm}(M(v_1)) - D^2_{gm}(M(v_1)) \right] \hat{n}_1$$
Finally, to descend this gradient, we establish a semi-implicit Euler scheme (see Supplemental Materials, section S1.3), with a step size parameter $\delta$, which we solve in the spectral domain as follows:

$$ u_{k+1} = F^{-1} \left\{ \frac{F\{\delta^{-1} u_{k} - g_{k}\}}{F\{\delta^{-1} + \alpha I - \beta \Delta \}} \right\}, $$

(12)

where $I$ denotes the identity operator.

### Implementation details, settings, and convergence.

The regseg tool includes a multiresolution strategy on the free-form deformation field. Registration pyramids are created by setting the spacing between the control points of the B-spline basis functions for each level of the multiresolution strategy. The implementation details as well as other features (such as the sparse matrix approach to fast interpolation) and the main parameters (such as $\delta$, $\alpha$, $\beta$, the B-spline grid resolutions, and target image smoothing) are discussed in Supplemental Materials, section S1. The actual choices of the parameter settings are publicly distributed with the source code for the experiments. The final settings were obtained manually based on feedback from the post-registration convergence reports (such as that found in Supplemental Materials, section S1.3). We released regseg along with the tool to generate the convergence report.

### 2.3. Evaluation protocol

In order to assess the performance of regseg, we defined the following general evaluation protocol: 1) Extract the set of undistorted surfaces $\Gamma_{r}$; 2) Compute a ground-truth field of displacements $U_{\text{true}}$, which is applied to generate warped images ($M$) for segmentation; 3) Execute regseg with $\Gamma_{r}$ and use the warped data as inputs; and 4) Perform a visual assessment and compute the error metrics. The adaptation of this protocol to the simulated phantoms and real data is explained in the following sections.

### 2.4. Simulated phantoms

The workflow required to simulate the digital phantoms and to assess the performance of regseg with them is presented in Figure 2. A set of four binary objects (i.e., “box”, “ball”, “L”, and “gyrus”) was generated by combining the binarization of analytical shapes and mathematical morphology. The reference surfaces $\Gamma_{r}$ were extracted from the binary shapes using FreeSurfer tools (Fischl, 2012). The ground-truth distortion was generated using a chain of two displacement fields supported by grids of B-spline basis functions. The coefficients of the basis functions were generated randomly for both levels in their three dimensions. The three components of the displacements $u = (u_{d})$ were bounded above by 40% of the separation between the control points at each level to obtain diffeomorphic transforms after concatenation (Rueckert et al., 2006). The first deformation field was applied to generate large warplings with control points separated by 50.50 mm in the three dimensions ($u_{d} \leq 20.20$ mm). With the second warping, we aimed to obtain a field with smoothness close to that found in a typical distortion field of dMRI data (Irfanoglu et al., 2011). Therefore, the control points were separated by 25.25 mm ($u_{d} \leq 10.10$ mm). After generating the ground-truth deformation, the original surfaces were warped by interpolating the displacements field at each vertex.

The warped surfaces $\Gamma_{\text{true}}$ were binarized to generate tissue fractions at low ($2.0 \times 2.0 \times 2.0$ [mm]) and high ($1.0 \times 1.0 \times 1.0$ [mm]) resolutions. Using a magnetic resonance (MR) simulator (Caruyer et al., 2014), we synthesized T1w (TE/TR= 10/1500 ms) and T2w images (TE/TR= 90/5000 ms), which corresponded to each phantom type, with each at two resolutions (1.0 mm and 2.0 mm isotropic). The field of view at both resolutions was $100 \times 100 \times 100$ [mm]. Next, regseg was applied to map $\Gamma_{r}$ onto the warped phantoms to obtain the registered surfaces ($\Gamma_{\text{test}}$). To quantify the misregistration error, we computed the Hausdorff distance between $\Gamma_{\text{test}}$ and $\Gamma_{\text{true}}$ using (Commandeur et al., 2011).
Figure 2: Evaluation of \textit{regseg} using phantom data according to the following instrumental workflow. 1) The reference surfaces $\Gamma_R$ are triangulated meshes extracted from the four binary shapes (i.e., “box”, “ball”, “L”, “gyrus”). 2) A ground-truth displacement field was generated as described in subsection 2.4, and applied to warp $\Gamma_R$, thereby obtaining $\Gamma_{true}$. 3) After being warped, $\Gamma_{true}$ were projected onto the corresponding discrete 3D volume and downsampled to create partial volume effects at two resolutions, i.e., $2.0 \times 2.0 \times 2.0$ [mm] and $1.0 \times 1.0 \times 1.0$ [mm], thereby producing sets of tissue fractions maps. 4) The tissue fractions were fed into a magnetic resonance (MR) simulator, which generated $T1w$ and $T2w$-like images at the two possible resolutions. 5) The \textit{regseg} tool was applied using the warped test images as multispectral moving images and $\Gamma_R$ as shape priors. 6) The agreement between the surfaces fitted by \textit{regseg} ($\Gamma_{est}$) and $\Gamma_{true}$ were assessed visually using automatically generated visual reports and quantitatively with the Hausdorff distance between the corresponding surfaces.

Figure 3: Experimental workflow employed to process real data from the Human Connectome Project (HCP). 1) $\Gamma_R$ were extracted from the anatomical reference (T1w image). 2) For use as the ground truth, we generated a plausible synthetic distortion $U_{true}$ from the field map with (13). 3) The dMRI data were warped using $U_{true}$ to reproduce the effects of real susceptibility-derived distortions. Target diffusion scalars (FA and ADC) were computed with the distorted data and stacked to feed the multivariate input required by \textit{regseg}. 4) The method was run to obtain $U_{est} = U_{true}$, i.e., the estimate of the ground-truth deformation. 5) The results were evaluated visually and quantitatively.
2.5. Real datasets

The general experimental framework for the real datasets is presented in Figure 3, which extends our previous evaluation (Esteban et al., 2014) of distortions using dMRI phantoms.

**Data.** To evaluate regseg using real dMRI data obtained from human brains, we collected 16 subjects from the Human Connectome Project (HCP) database. The original acquisitions are released within “unprocessed” packages, whereas the “minimally preprocessed” packages contain the corresponding images after some processing (correction for several artifacts, brain-extraction, spatial normalization, etc.). We refer the reader to (Van Essen et al., 2012) for exact details of the acquisition parameters and (Glasser et al., 2013) for the preprocessing issues. These datasets comprise a large set of images, including T1w, T2w, and multi-shell dMRI images.

**Segmentation model.** Based on our experience and previous studies (Ennis and Kindlmann, 2006), we defined the moving image as a stack of the FA and ADC maps derived from dMRI data. After evaluating several alternative models, we empirically defined a partition \( \Omega \) according to the following six regions: 1) thalamus \( (\Omega_{Tha}) \); 2) ventricular system and deep GM structures \( (\Omega_{VdGM}) \); 3) cerebral WM \( (\Omega_{WM}) \); 4) brain stem and cerebellar WM \( (\Omega_{bst}) \); 5) cerebellar GM \( (\Omega_{cbGM}) \); and 6) cortical GM \( (\Omega_{GM}) \). Using tools in FreeSurfer and appropriate selections of labels in the aparc segmentation released with the HCP data, we extracted the \( \Gamma_R \) set for the reference surfaces. The segmentation model corresponding to this partition is shown in Figure 4 and discussed in greater detail in Supplemental Materials, section S4.

**Ground-truth generation.** Realistic deformation is achieved by generating displacements fields that satisfy the theoretical properties of distortion (Jezzard and Balaban, 1995). The displacements along the phase-encoding (PE) axis of the dMRI image are related to the local deviation of the field \( \Delta B_0(r) \) from its nominal value \( B_0 \), as follows:

\[
u_{PE} = \frac{\gamma T_{acq} s_{PE}}{2\pi} \Delta B_0(r) \text{[mm]}, \tag{13}\]

where \( \gamma \) is the gyromagnetic ratio, \( T_{acq} \) is the readout time, and \( s_{PE} \) is the pixel size along PE. Certain MR sequences are designed to estimate \( \Delta B_0 \), thereby obtaining the so-called field map. We derived the deformation \( U_{true} \) from the field map image released with the corresponding packages of each dataset in the HCP. The field map was unwrapped\(^2\) and smoothed before applying (13). Next, the original dMRI was warped using the resulting displacement field and fed into a pipeline to process the corresponding diffusion tensor image (DTI), thereby computing the derived scalars of interest (FA and ADC) using MRtrix (Tournier et al., 2012).

**Metric assessment.** Initially, we investigated the appropriateness of the segmentation model. For five test datasets, we uniformly sampled the space of distortions \( U = \epsilon \cdot U_{true} = r + \epsilon u_{PE} \) (with \( \epsilon \in [-1.1, 1.1] \)) and evaluated the data term of the cost function (7). The minimum of the cost function (subsection 2.1) was consistently located at \( \epsilon = 0.0 \) (the ground-truth) for all of the cases (Supplemental Materials, Figure S2).

**Cross-comparison.** A similar workflow to the general evaluation framework used for Figure 3 was employed to integrate the alternate T2B registration scheme. We reproduced the solution and settings provided with ExploreDTI (Leemans et al., 2009), which is a widely used toolkit for tractography analysis of DTI. ExploreDTI internally employs elastix (Klein et al., 2010) to perform registration.

\(^2\)Fieldmaps are phase maps, which are intrinsically clipped in the interval of \([-\pi, \pi]\) [rads] or [rads/s].
Figure 4: Segmentation model defined by the homogeneous regions $\Omega_i$, where the image for segmentation is partitioned by the theoretical projection ($\Gamma_{\text{true}}$) of $\Gamma_R$ onto $M$. The plot represents a kernel density estimate of the location and spread of each region $\Omega_i$ in the bivariate feature space of $M$, which comprises the FA and ADC maps. At the top and the right margins, the marginal distributions of $\Omega_i$ are also plotted for the FA and ADC, respectively.
1. Error measurement. Distortion only occurs along the PE axis of the image, so we computed the surface warping index (sWI) as the area-weighted distance between the corresponding vertices of $\Gamma_{true}$ and their estimate obtained by the method under the test $\hat{\Gamma}_{test}$:

$$sWI = \frac{1}{\sum_i a_i} \sum_i a_i \|v_i - \hat{v}_i\|,$$

(14)

where $v_i \subset \Gamma_{true}$ are the locations of the total $N_V$ vertices and $\hat{v}_i \subset \hat{\Gamma}_{test}$ are the recovered locations that correspond to $v_i$. In practice, we only report the sWI for three surfaces of crucial interest in whole-brain tractography. These three surfaces delineated the following regions in the model: $\Omega_{VdGM}$, $\Omega_{WM}$, and $\Omega_{GM}$.

3. Results

3.1. Proof of concept using digital phantoms

In total, 1200 experiments (four phantom types $\times$ 150 random warpings $\times$ two resolutions) were performed according to the workflow illustrated in Figure 2. For each experiment, the misregistration error was measured using the Hausdorff distance (see section 2.5) between the theoretical $\Gamma_{true}$ and the estimate obtained by regseg ($\hat{\Gamma}_{test}$). The results demonstrated that the accuracy was consistent and high, and below the image resolution. Figure 5 (block C) shows the violin plots for each model type corresponding to the two sets of resolutions for the generated phantoms. In order to relate the average misregistration error to the resolution of the moving image, we proceeded as follows. First, we confirmed that the vertex-wise error distributions were skewed by using Shapiro-Wilk’s test of normality. All of the distributions of errors in the tests (four phantom types $\times$ two resolutions) were non-normal with $p < 0.001$. Consequently, we used the non-parametric Wilcoxon signed-rank test with the Bonferroni correction for multiple comparisons ($N = 150$, for each phantom type). The average errors were significantly lower than the voxel size with $p < (0.001/150)$ in all tests (four phantom types $\times$ two resolutions). Statistical tests might not be sufficiently conclusive, so we also computed the confidence intervals, as shown in Table 1.

3.2. Evaluation using real datasets and cross-comparison

Finally, we compared the performance of regseg with that of the standard T2B method. Summary reports for visual assessment of the 16 cases are included in Supplemental Materials, section S5. In Figure 6, box A, the visual report is shown for one subject. We computed the sWI (14) of every surface after registration using both the regseg and T2B methods. Finally, to compare the results, we performed Kruskal-Wallis H-tests (a non-parametric alternative to ANOVA) on the warping indices for the three surfaces of interest selected in section 2.5 ($\Gamma_{VdGM}$, $\Gamma_{WM}$, $\Gamma_{pial}$). All of the statistical tests showed that the error distributions obtained with regseg and T2B were significantly different, and the violin plots in box B of Figure 6 demonstrate that the errors were always larger with T2B. We also show the 95% CIs of the sWI for these surfaces. The aggregate CI for regseg was 0.56–0.66 [mm], whereas the T2B method yielded an aggregate CI of 2.05–2.39 [mm]. The results of the statistical tests and the CIs are summarized in Table 2.
Figure 5: A. Visual assessment of the results obtained with the low resolution sets: “gyrus” (top left), “L” (top right), “ball” (bottom left), and “box” at (bottom right). The contours recovered after registration are represented in yellow. Regseg achieved high accuracy because it determined the almost exact locations of the registered contours with respect to their ground truth positions (shown in green). The partial volume effect makes segmentation of the sulci a challenging problem with voxel-wise clustering methods, but they were successfully segmented with regseg. B. Quantitative evaluation of registration errors in terms of the average Hausdorff distances between surfaces at low (left) and high (right) resolutions, which demonstrate that the errors were consistently below the size of the voxels.

Table 2: Statistical analysis of results obtained using real data, which show that regseg performed better than the alternative T2w-registration based (T2B) method. The distribution of the errors computed for the surfaces of interest ($\Gamma_{V,AGM}$, $\Gamma_{W,M}$, $\Gamma_{pial}$) and the aggregate of all surfaces (Aggreg. column) had lower 95% CIs with regseg. The CIs in this table were computed by bootstrapping using the mean as the location statistic and with $10^4$ samples. The Kruskal-Wallis H-tests indicated that there was a significant difference between the results obtained using regseg and the T2B method.
Figure 6: A. Example of a visual assessment report, which was generated automatically by the evaluation tool. Each view shows one component of the input image (in this case, the FA map), the ground-truth locations of the surfaces (green contours), and the resulting surfaces obtained with the test method (yellow contours). The first two rows show axial slices for regseg and the T2w-registration based (T2B) method, while the last two rows show the corresponding sagittal views. The coronal view is omitted because it was the least informative due to the directional property of the distortions. Specific regions where regseg outperformed T2B are enlarged. B. Violin plots of the error distributions for each surface, which show the voxel size of the dMRI images (1.25 mm), thereby supporting the improved results obtained by regseg with the proposed settings.
4. Discussion

Accuracy tests. The hypothesis tested in our study was that reliable image registration can be performed by searching for homogeneous regions in a multispectral image, which correspond to precise contours from an atlas, or extracted from another image (i.e., a different time step). We demonstrated that active contours without edges can be used successfully to drive a deformation supported by B-spline basis functions with digital phantoms. We randomly deformed four different phantom models to mimic three homogeneous regions (WM, GM, and cerebrospinal fluid) and we used them to simulate T1w and T2w images at two resolution levels. After registration with regseg, we measured the Hausdorff distance between the projected contours obtained using the ground-truth warping and our estimates. We concluded that the errors were significantly lower than the voxel resolution. We also assessed the 95% confidence interval (CI), which yielded an aggregate interval of 0.64–0.66 [mm] for the low resolution phantoms (2.0 mm isotropic voxel) and 0.34–0.38 [mm] for the high resolution phantoms (1.0 mm isotropic). Therefore, we also concluded that the error was bounded above by half of the voxel spacing.

Application to real data. We designed regseg as a method for segmenting dMRI data by exploiting the anatomical information extracted from the corresponding T1w image of the subject. Applications in whole-brain tractography (Smith et al., 2006; Craddock et al., 2013) usually solve this problem with a two-step approach. First, the images are corrected for nonlinear distortions using auxiliary acquisitions such as field maps (Jezzard and Balaban, 1995), T2w images (Kybic et al., 2000). Second, the segmentation is projected from a reference T1w image using linear registration (Greve and Fischl, 2009). Regseg addresses this joint problem in a single step and it does not require any additional acquisition other than the minimal protocol using only T1w and dMRI images. This situation is found commonly in historical datasets.

We evaluated regseg in a real environment using the experimental framework presented in Figure 3. We processed 16 subjects from the HCP database using both regseg and an in-house replication of the T2w-registration based (T2B) method. Regseg obtained very high accuracy, with an aggregate 95% CI of 0.56–0.66 [mm], which was below the pixel size of 1.25 mm. The misregistration error that remained after regseg was significantly lower ($p < 0.01$) than the error corresponding to the T2B correction according to Kruskal-Wallis H-tests (Table 2). Visual inspections of all the results (Supplemental Materials, section S5) and the violin plots in Figure 6 confirmed that regseg performed better than the T2B method in our settings. We carefully configured the T2B method using the same algorithm and the same settings employed in a widely-used tool. However, cross-comparison experiments are prone to so-called instrumentation bias (Tustison et al., 2013). Therefore, these results do not prove that regseg is better than T2B. Our results suggest that regseg is a reliable option in this application field. In addition, the T2B may introduce an additional (and small) error during the necessary registration of T2w in the T1w space.

Conclusion

Regseg is a variational framework for the simultaneous segmentation and registration of 3D dMRI data obtained from the human brain, where within-subject anatomical information is used as a reference. The registration method segments the target multivariate image into several competing regions, which are defined explicitly by their limiting surfaces. The surfaces are active and they evolve on a free-form deformation field supported by the B-spline basis. A descent optimization strategy is guided by shape gradients computed on the current partition of the target image. Regseg uses active contours without edges and it searches for homogeneous regions within the image. We tested regseg using digital phantoms by simulating T1w and T2w magnetic resonance imaging (MRI) warped with smooth random deformations. The resulting misregistration of the contours was significantly lower than the image resolution of the phantoms.

We proposed regseg for simultaneously segmenting and registering dMRI data to their corresponding T1w image from the same subject. We demonstrated the accuracy of the proposed method based on visual assessments of the results obtained by regseg and cross-comparisons with a widely used technique. Moreover, regseg does not require any images in addition to the minimal acquisition protocol, which only utilizes T1w and dMRI. As well as the proposed application to dMRI data, other potential uses of regseg are atlas-based segmentation and tracking objects in time-series.
Availability and reproducibility statement

We considered the reproducibility of our results as a design requirement. Therefore, we used real data obtained from a publicly available repository (the Human Connectome Project (Van Essen et al., 2012)) and all of the software utilized in this study is also publicly available. Regseg was implemented on top of ITK-4.6 (Insight Registration and Segmentation Toolkit, http://www.itk.org). The evaluation instruments (Figure 3) were implemented using nipype (Gorgolewski et al., 2011) to assess their reproducibility. All of the research elements (data, source code, figures, manuscript sources, etc.) involved in this study are publicly available under a unique package (Esteban and Zosso, 2015).

Author contributions

All the authors contributed to this study. OE implemented the method, designed and conducted the experiments, wrote the paper, simulated the phantoms, and prepared the real data. DZ devised and drafted the registration method, generated early phantom datasets, and collaborated in the implementation of the method. AD, MBC, and MJLC interpreted the results. AD, MBC, MJLC, JPT, and AS advised on all aspects of the study.

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Appendix

Appendix 1. Simplifying the regularization term

The exponentials of the Thikonov regularization prior (6) have the general form $v^T M v$. If $M$ is a $n \times n$ diagonal matrix such that $M = \mathbf{m} \mathbf{I}_n$, then:

$$v^T M v = \mathbf{m} \cdot (v^T \mathbf{I}_n v) = \mathbf{m} \cdot v^\circ 2,$$

where we have introduced the Hadamard power notation$^3$.

In general, the anisotropy is aligned with the imaging axes, so $A$ and $B$ of (7) can be simplified to diagonal matrices, such that $A = \alpha \mathbf{I}_n$ and $B = \beta \mathbf{I}_n$. By substituting into equation (7), we obtain:

$$E(u) = \text{Const.} + \sum_i \int_{\Omega_i} D_i^2(f') \, dr + \int_{\Omega} \frac{1}{2} [\alpha \cdot u^\circ 2 + \beta \cdot (\nabla u)^\circ 2] \, dr.$$  \hspace{1cm} (A.1)

$^3$The Hadamard power of a matrix or a vector is the power of its elements $M^{\circ p} = (m_{ij}^p)$.


