A Local Phase-Based Algorithm for Registration of CMR Scans from Multiple Visits

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Abstract

A variational 3D phase-based registration approach for the alignment of reconstructed cine volumes from short- and long-axis cine CMR slices is proposed. The algorithm performance is validated comparatively with an equivalent intensity-based approach on an artificially generated phantom volume with a pre-prescribed intensity difference. Final results are shown for the registration of clinical cine CMR volumetric alignment.

1. Introduction

The provision of both anatomical and functional data relating to healthy and pathological cardiac tissue by cardiovascular magnetic resonance (CMR) imaging has been invaluable, enabling earlier diagnosis and improvement in effective treatment of ischaemic heart disease.

In patients who have suffered cardiac injury, such as in cases of myocardial ischaemia, longitudinal analysis of the diseased tissue composition provides important information regarding left ventricular (LV) remodelling. To perform such longitudinal analysis, it is necessary to establish a correspondence between tissues imaged on separate occasions. Conventional 2D registration of clinical CMR images acquired at different times has been used for the comparison of imaging findings. However, it is well known, and we have previously shown [1], that discrepancies exist between image planes selected on separate hospital visits, resulting in significant out-of-slice displacements which cannot be compensated by 2D registration.

As a result of these observations, in this paper, we propose a 3D registration approach for longitudinal CMR analysis based on a variational optical flow framework. Moreover, we propose the use of a registration approach whose similarity metric is based on local phase rather than image intensities. We propose such an approach because of the non-quantitative relationship between tissue characteristics and image intensity in CMR, as a result of which corresponding tissues imaged on separate hospital visits may have significantly different intensities. Local phase is a contrast independent descriptor of image structure and is thus not affected by such intensity inconsistencies.

A large amount of literature exists relating to the registration of cardiac data – for reviews see [2,3]. In relation to our own work we note [4,5] where optical flow methodologies have been used for cardiac MR registration, whilst phase-based similarity metrics have been previously looked at in [6,7] in the context of cardiac MR data alignment.

2. Materials and methods

2.1. Artificial data

A synthetic phantom image, representing the left (LV) and right (RV) ventricles as well as the myocardium of an idealised heart was created. The phantom provides data with a known ground truth for validation purposes. In all validation tests Gaussian white noise was added to the original phantom image.

2.2. Clinical data

Image datasets were acquired using a SIEMENS TrioTim 3 Tesla scanner at the John Radcliffe Hospital, Oxford. For each subject two studies were acquired, the second obtained 6 months after the first. Each data set consisted of on average 3 long-axis (LA) images and a short-axis (SA) cine stack with 8 to 10 slices. The SA slices had pixel sizes of approximately 1.56mm x 1.56mm x 8mm. The SA stack slices were acquired at regularly spaced (10mm) locations along the LA orientation. The pixel size in LA images was approximately 1.41mm x 1.41mm x 8mm. The relative positions of all SA and LA slices in the datasets were available via the DICOM header. In addition to the cine scans, other CMR scans including late gadolinium enhancement images were acquired as part of the protocol. However, the work presented in this paper deals only with the cine scans.

In order to perform 3D registration, volumetric reconstruction must first be undertaken using information from both the SA and LA cine MR data. Intensity differences between SA and LA data require the use of an intensity correction algorithm. The intensities in the LA are corrected based on methods presented in [8]. The first and second order intensity moments at the SA-LA intersection points are matched and these corrections propagated to neighbouring LA pixels based on intensity similarities. From the newly corrected image data isotropic volumetric representations of the imaged cardiac tissue may be generated.

2.3. Registration framework

The classical optical flow registration approach is based on an intensity constancy assumption, formulated as in equation 1.

$$I(x, y, z, t) = I(x + dx, y + dy, z + dz, t + dt) \qquad (1)$$

$$I_x v_x + I_y v_y + I_z v_z = -I_t \qquad (2)$$

Linearization by Taylor's expansion gives the optical flow constraint equation written in equation 2 where I_t is approximated by the difference between the two images, I_x, I_y, and I_z are oriented differentials of the image and v_x, v_y, and v_z compose the unknown deformation field between the two images.

The ill-posed nature of the optical flow constraint equation requires further constraint in order to find a unique solution. In [9] a methodology is proposed which enforces the global smoothness of the deformation field allowing a solution via the the corresponding Euler-Lagrange equations. A local approach is adopted in [10] where the assumption of flow uniformity in local neighbourhoods enables calculation of a unique solution. There are known disadvantages to both local and global approaches however, those being the sparsity of the final solution and a higher susceptibility to image noise respectively.

A desirable algorithm is therefore one that combines the advantages of both local and global methods. Such a method was presented by Bruhn and Weickert in [11]. This is the framework utilized in our proposed approach, however rather than base our similarity metric on intensity values as in the original paper we use the difference of local phase values as our metric.

2.4. Local phase calculation

As previously mentioned there is no guarantee of intensity constancy between cine MR images acquired during different imaging sessions. As a result metrics based on the image intensities may not lead to satisfactory registration solutions.

In monomodal registration we expect the images to contain the same tissue structures; therefore a method which utilizes this shared information is desirable. A viable option in such cases is local phase which can be used as a brightness and contrast invariant descriptor of image structure.

When considering a 1D signal, local phase can be calculated from the analytic signal. A number of approaches have been proposed to extend the concept of the analytic signal to higher dimensions. Two commonly used are the application of multiple directional quadrature filters, which enable the approximation of motion along the given orientation, or alternatively the monogenic signal.

In the work presented here we utilized 9 oriented filters as proposed in [12]. Our local phase values replace the intensity values that were previously used in Bruhn-Weickert's local-global registration framework. The form of the modified registration framework is shown in equations 3 and 4. In equation 3 α is used to control the influence of the local and global components to the final deformation field solution, whilst in equation 4 the 9 local phase (φ) responses to the directional filters are combined, before applying a Gaussian filter of variance ρ, denoted by h_ρ, for local smoothing.

The final part of the registration algorithm is the inclusion of a confidence measure which gives higher weights to some phase values in the images relative to others, denoted by C_n in equation 4. The confidence measure used was utilized by Hemmendorff in [12]. The aim of the confidence measure is twofold: to eradicate any phase singularities occurring due to interfering frequencies and to increase the influence of strong edges over weak ones. The confidence measure is composed of 4 terms based on the filter output magnitude, the phase linearity, the phase gradient similarity and the frequency reliability.

$$E = \iiint \left(w J_\rho w^T + \alpha \|\nabla w\|^2 \right) dx\, dy\, dz \qquad (3)$$

$$where\ w = [v_x, v_y, v_z, 1]^T$$

$$J_\rho = h_\rho * \sum_{n=1}^{9} \frac{C_n{}^2}{9} \begin{pmatrix} \varphi_{x,n}{}^2 & \varphi_{x,n}\varphi_{y,n} & \varphi_{x,n}\varphi_{z,n} & \varphi_{x,n}\varphi_{t,n} \\ \varphi_{y,n}\varphi_{x,n} & \varphi_{y,n}{}^2 & \varphi_{y,n}\varphi_{z,n} & \varphi_{y,n}\varphi_{t,n} \\ \varphi_{z,n}\varphi_{x,n} & \varphi_{z,n}\varphi_{y,n} & \varphi_{z,n}{}^2 & \varphi_{z,n}\varphi_{t,n} \\ \varphi_{t,n}\varphi_{x,n} & \varphi_{t,n}\varphi_{y,n} & \varphi_{t,n}\varphi_{z,n} & \varphi_{t,n}{}^2 \end{pmatrix} \quad where\ n = 1{:}9\ (filter\ directions) \qquad (4)$$

3. Results

3.1. Phantom

The known segmentations of the LV, RV and myocardium in our phantom data set enable the calculation of overlap metrics to determine the relative accuracy of the registration algorithm under different conditions. A range of displacement and angular offsets were applied to the phantom and the results of registration quantified in terms of overlap accuracy using the Dice coefficient (DC). The offsets were varied in x, y, and z between 0mm and 6mm, whilst the angular change was varied between 0^o and 12^o. The mean Dice coefficient, taking into account all combinations of offsets, were 0.9614 ± 0.0202, 0.8916 ± 0.0543 and 0.8160 ± 0.0830 for the LV, RV and myocardium respectively.

A second test was performed to compare the results obtained using our phase-based implementation with results obtained using the equivalent intensity-based implementation, where the derivatives of phase in equation 4 are replaced by the intensity derivatives. Our hypothesis is that our phase-based method copes better with potential intensity variations between the images being aligned. For this reason the test carried out involved

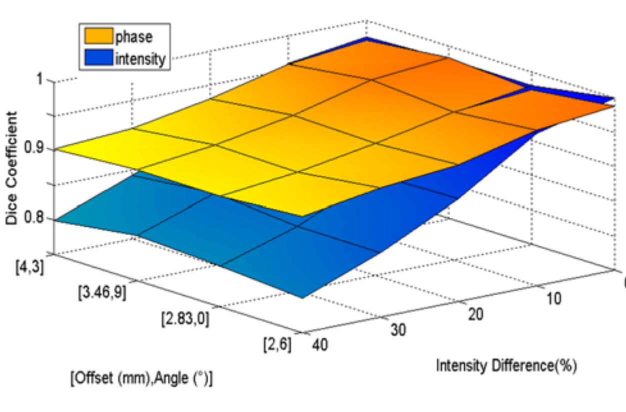

Figure 1. Dice coefficients calculated for both phase- and intensity-based registration over varying intensity differences.

a small selection of offsets tested with a range of intensity differences applied between the images. The intensities of each image component (the LV, RV and myocardium) were varied such that the absolute intensity differences between corresponding structures in the two volumes took values starting at 0% up to a maximum difference of 40%. The Dice coefficient results for the LV are shown in figure 1. The results for the RV and myocardium followed similar trends.

3.2 Clinical data

Registration was performed on reconstructed cine volumes from patients who underwent CMR imaging on two separate visits to the hospital. Example results are shown in figure 2.

4. Discussion

A local phase-based 3D registration method was proposed which combines the benefits of global and local methods. The use of phase increases the robustness of the algorithm to noise and intensity differences between images.

1) Phantom Study: The algorithm was first tested on a phantom image where initial results were promising, achieving reasonable Dice coefficients consistently over a range of offset values. Our hypothesis that our phase-based method would be more able to deal with intensity differences than an intensity-based method was also confirmed by the results shown in figure 1. We see that beyond 10% intensity difference our phase-based method consistently performs better.

2) MRI Study: Results for real data shown in figure 2 all show successful alignment of longitudinal studies. Continuation of contours can be clearly seen in each case. Whilst these preliminary results are promising, further validation using quantitative measures of accuracy needs to be undertaken. A second limitation to the current work is that it requires a reasonably close initialisation. It may be necessary therefore in future work to look at initialisation registration methods that use either a rigid or affine motion model.

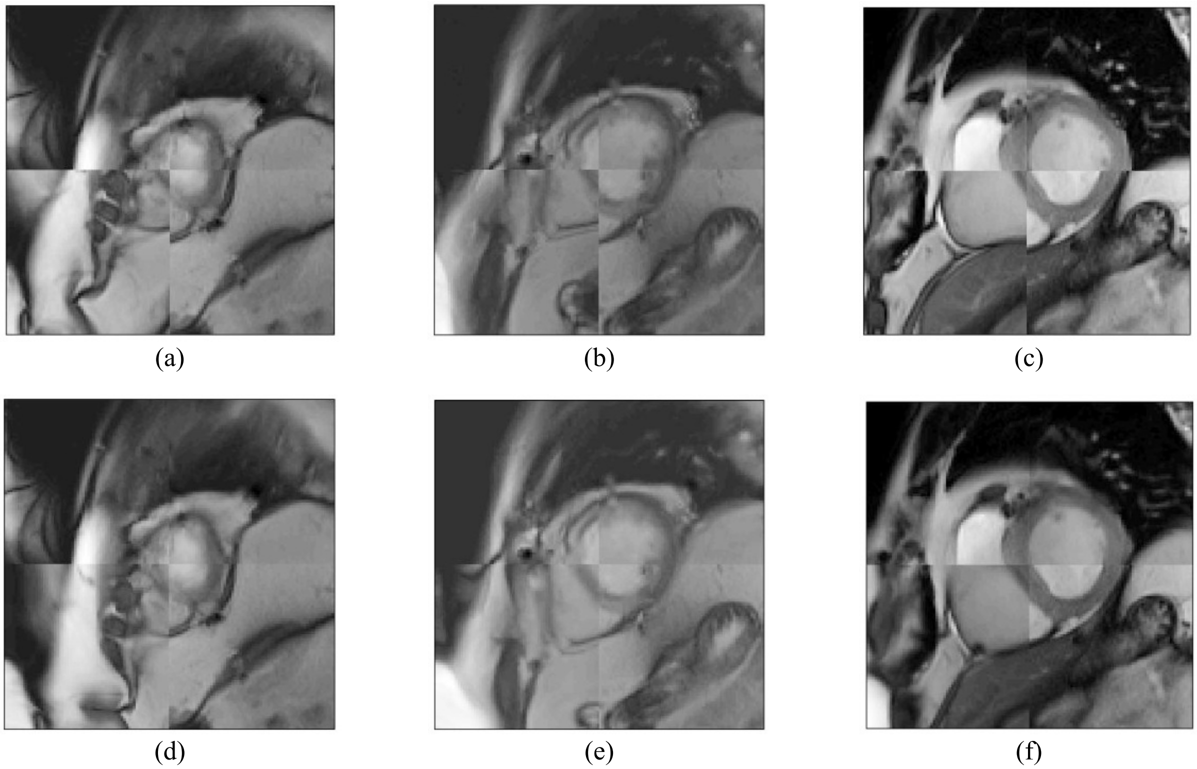

Figure 2. (a)→(c) show the original images in a checkerboard pattern (top left and bottom right quadrants from one visit, bottom left and top right from the other). (d)→(f) show the same slices after registration.

Acknowledgements

The research was funded by an EPSRC DTA studentship.

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