Implementation of Displacement Field Fitting for Calculating 3D Myocardial Deformations from Parallel-Tagged MR Images

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Abstract

The utility of using a displacement field fitting approach to reconstruct 3D myocardial deformations from sets of parallel-tagged MR images has been demonstrated [1]. Our present goal is to implement the field fitting algorithm efficiently on real heart data. A cascade of fitting steps is used: a first order global cartesian fit, followed by an n-th order global prolate spheroidal fit, followed by a local prolate spheroidal fit around each pre-selected test point. Local deformation gradient and strain tensors are then computed by taking directional derivatives of the displacement field at each test point. On a normal human volunteer, the standard deviation of error of the fitting model to the raw tag point displacement data was ~0.3mm. Projections of representative image slices following each stage in the cascade provide a graphical validation of the technique.

Introduction

Cardiac dysfunction remains the number one cause of mortality in this country. The techniques of MRI tagging [2] and fast, breath-hold imaging [3] have emerged as promising methods for studying the function of the heart wall non-invasively in both the research and clinical settings. The objective of this research is to non-invasively describe of the time evolution of the three dimensional (3D) deformation field within the left ventricular wall. A typical MRI tag data set is generated using three sets of image slices, each spanning the entire volume of the heart. Myocardial tags are regions of the sample where the spins of the hydrogen nuclei have been saturated, producing a signal void when subsequently imaged. Since the tags result from perturbations of the tissue itself, the deformation of the tags accurately reflect the motion of the underlying tissue [4]. In any image set, a series of tag planes oriented along one of the three orthogonal coordinate axes and perpendicular to the image set are prescribed using local magnetic field gradients and radio frequency pulses. Images sets are then generated at 35ms intervals throughout the contraction phase of the heart cycle. Because the MRI data is acquired in fixed, 2D cross-sections, special techniques are needed to reconstruct the 3D heart motion.

The 3D reconstruction technique we developed, displacement field fitting, considers points on the imaged tags as samples on the surfaces of the deformed tag planes. The 1D displacement of points from their initial tag plane locations can be used to fit for the coefficients in an expression describing the displacement field in that particular direction. After fitting for the 3 independent orthogonal displacement fields, the interpolation expressions can be combined to track any point in the heart in 3D. Using a prolate spheroidal coordinate system has been shown to increase the accuracy of the fit with fewer fitting parameters [1]. To assist in the tracking of the prolate spheroidal centroid and focal point and to account for gross deformations of the heart, a low order fit in cartesian coordinates is performed over the entire heart data set (globally) as an initial step. Then, to account for the expected large-scale motions of the heart (including transmural twist, axial torsion and differential thickening from the endo- to epicardium), a global fit in prolate spheroidal coordinates is performed. Finally, to account for local perturbations in the deformation field, a local (or small region of interest) fit in prolate spheroidal coordinates is performed around each of a predetermined set of test points regularly spaced within and around the heart.

Methods

Data from a typical 3D human data set at 228ms after the QRS wave are shown in figure 1. The cartesian coordinate fitting expansion was defined simply as a 1st order power series in (x,y,z) for each of the three cartesian displacement directions: e.g. $\delta x = a_0 + a_1 x + a_2 y + a_3 z$. The unknown coefficients (a_0, a_1, a_2, a_3) were solved using the measured MRI point displacement data values via singular value decomposition (a least squares fitting method). To define the prolate spheroidal coordinate system centroid and apical focal point, a prolate spheroid of constant radius (λ) was fitted to the LV epicardial surface points. A prolate spheroidal expansion was then contructed and fitted to the tag displacements in a manner analogous to that used for the cartesian fitting expansion, but using series terms similar to spherical harmonics.

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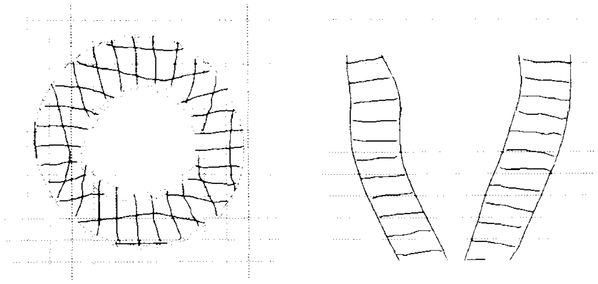

Figure 1: Short axis (Left) and long axis image slices in the normal human heart at 228ms into systole. Two short axis image sets are superimposed giving a grid tag appearance. Undeformed tag plane locations are also shown.

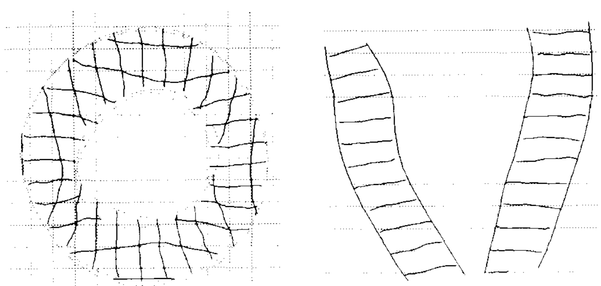

Figure 2: Image slices after correction for bulk motions and linear cartesian shears and stretches.

The 3D boundaries of the endo- and epi- cardial LV surfaces were fit to a prolate spheroidal expansion of λ as a function of θ and ϕ and a set of test points, regularly spaced in λ, ϕ and θ, were defined within and around the myocardium. Finally, at each test point a 1st order prolate spheroidal fit was computed from data in a localized region around each point. The region of interest (ROI) size was defined as the distance between adjacent short axis in a given short axis set, and the angular distance between adjacent long axis slices. A five pole butterworth filter over the ROI limits was used to reduce discontinues in neighboring fits.

Results

The standard deviation in the error (SDE) of the fitting model to the raw tag point displacement data after the initial 1st order cartesian fitting step was 1.63mm. The full 3D tag data set was then undeformed accordingly. This is shown in figure 2. A 1st order radial, 4th order angular, prolate spheroidal fit was then performed on the residual displacement values. The overall SDE of the fit to the data was reduced to 0.41mm and the corrected data set is shown in figure 3. A 6th order prolate spheroidal surface fit was then performed and 96 test points defined. The average SDE of the fit around each test point was 0.314mm. The final corrected data are shown in figure 4.

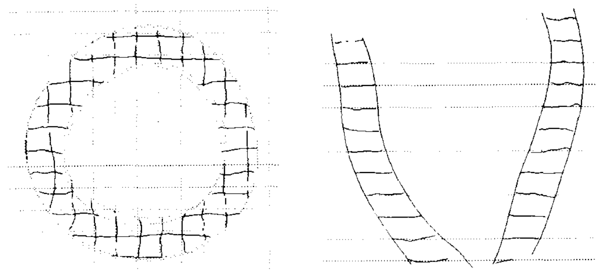

Figure 3: After correction for global prolate spheroidal motions.

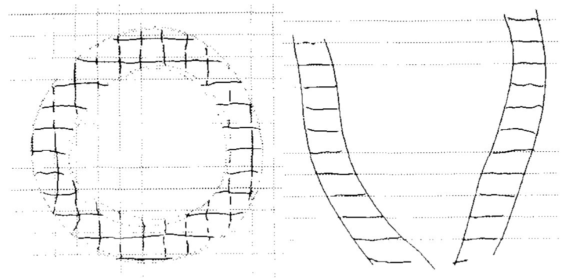

Figure 4: After correction for local linear prolate spheroidal fitting.

Conclusion

The displacement field fitting method outlined here achieved ~0.3mm fitting error for a normal human MRI data set using a 1st order global cartesian fit (12 fitting parameters), a 150 parameter global prolate spheroidal fit and a 1st order local prolate spheroidal fit (24 fitting parameters). This error approaches the expected 0.1–0.2mm error in the determination of the tag point displacements from clinical MR parallel tag data. This is also in the range of the uncertainties in the 3D position tracking of radio-opaque markers in cine X-ray film studies, the previous gold standard for 3D strain calculations in intact heart preparations.

References

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