

Improving HARP Cardiac Strain Mapping Using Nonlinear Diffusion

K. Z. Abd-Elmoniem¹, V. Parthasarathy¹, J. L. Prince¹

¹Johns Hopkins University, Baltimore, MD, United States

Introduction: Harmonic phase (HARP) MRI is used to measure myocardial motion and strain from tagged MR images (1). HARP MRI uses a limited number of samples from the spectrum of the tagged images to reconstruct motion and strain. The HARP strain maps, however, suffer from artifacts that limit the accuracy of the computations and degrade the appearance of the strain maps. Causes of these so-called “zebra” artifacts include image noise, Gibbs ringing, and interference from other Fourier spectral peaks. Computing derivatives of the HARP phase, which are needed to estimate strain, further accentuates these artifacts. Previous methods to reduce these artifacts include 1-D (2) and 2-D nonlinear filtering of the HARP derivatives (3), and a 2-D linear filtering of unwrapped HARP phase (4). A common drawback among these methods is the lack of proper segmentation of the myocardium from the blood pool. The lack of segmentation allows noisy phase values from the blood pool to be incorporated in the smoothed strain maps, which produces artifacts. In this work, we propose a smoothing method based on anisotropic diffusion that filters the HARP derivatives strictly within the myocardium without the need for prior segmentation. The information about tissue geometry and the strain distribution is used to restrict the smoothing to within the myocardium, thereby ensuring minimum distortion of the final strain map. Preliminary results demonstrate the ability of anisotropic diffusion to reduce artifacts and improve strain accuracy over existing methods.

Theory and Methods: Nonlinear anisotropic diffusion (NLD) is a partial differential equation (PDE)-based regularization algorithm for smoothing an image while preserving the sharpness of boundaries and edges. It forces smoothing to occur along directions of minimal intensity variations, i.e. directions orthogonal to edge normals (5). This framework was recently extended to multi-valued images with curvature preservation. We model the combination of harmonic magnitude and gradients of HARP phase as a multi-valued image $S(x,t) = [D(x,t), \nabla_{x_1} u(x,t), \nabla_{x_2} u(x,t)]$, where $D(x,t)$ is the structure tensor field and is computed using the HARP magnitude while $\nabla_{x_2} u$ and $\nabla_{x_1} u$ are the gradient fields of the 2-D displacement which serve in our framework as the vector-valued intensity map. Constructing such a multi-valued image prevents smoothing both across edges of the myocardium and also across edges in displacement. The preservation of edges in displacement ensures that the subtle changes in strain are not blurred out. The vector-valued S is smoothed using the PDE $\partial S_i / \partial t = \text{div}(T \nabla S_i)$, $\forall i = 1, 2, 3$, where $T(S)$ is the image dependent diffusion tensor that drives the regularization process.

Simulation and Experiments: Two phantom simulations of healthy and diseased cardiac states and one in-vivo experiments were conducted in this study. The healthy state simulation is created using an incompressible, contracting myocardium, shaped like an annulus with an inner radius $R_i = 22.5$ mm and outer radius $R_o = 29.5$ mm, that deforms under radially-varying contraction. The maximum radial strain, E_r , occurring in the myocardium is 18% and the maximum circumferential strain, E_{cc} , is -21%. Maximum strains occur at the endocardium. FOV=280 mm and tag separation = 7 mm. Noise was added to the tagged image with a contrast-to-noise ratio of 15. A HARP bandpass filter size of 72 was used to compute the circumferential and radial strains. In the diseased state simulation, an asymmetric strain pattern was created with the same initial radii with FOV= 320 mm and tag separation = 8 mm. E_r ranges from 2% to 16% and E_{cc} ranges from -26% to -8%. A healthy adult subject scan was also conducted with the parameters: acquisition window: 15ms, spiral interleaves: 12, res. 256x256, FOV 350mm, TE 4.0ms, TR 30ms, slice thickness 6mm, tag-spacing 8mm.

Results and Discussion: Fig. 1 shows a visual comparison of the E_{cc} of the different smoothing methods. The left panel shows the simulated healthy set. We see that the zebra artifact seen in the no-smoothing case does not disappear when median and the phase unwrapping methods are used. The 1-D smoothing reduces the artifacts, but introduces gross errors (notice the loss of dark blue coloration corresponding to maximum circumferential strain). On the other hand, the anisotropic diffusion reduces the artifact without introducing gross errors. The right panel shows the simulated diseased set in which the errors are more severe. While the 1-D method and the median method do not seem to improve the appearance much, the weighted phase unwrapping method introduces large errors in endocardium. Using anisotropic diffusion, we see a considerable improvement in the appearance of the map. A quantitative comparison is given in Fig. 2. In nearly all cases, the error is smallest when using anisotropic diffusion (black curve). Fig. 3 shows the in-vivo E_{cc} strain maps at different cardiac phases both with and without the proposed nonlinear smoothing scheme. Ringing artifacts are reduced at different myocardial regions (note the area marked using arrows) and the overall visual quality of the strain maps improves considerably.

Conclusion: A smoothing method based on anisotropic diffusion is proposed to smooth HARP strain maps. Using simulations and in-vivo experiments, it is shown that the algorithm improves the accuracy of the strain maps and also improves their visual quality. We believe that the proposed method will improve the diagnostic usability of HARP strain maps.

References:

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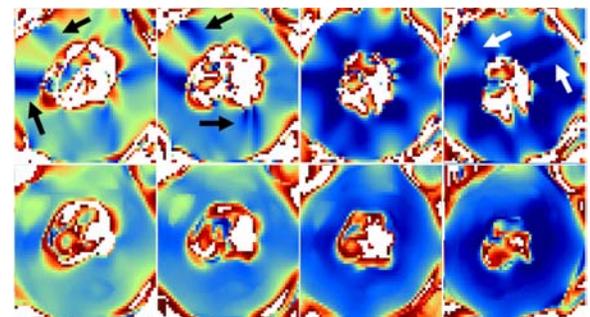
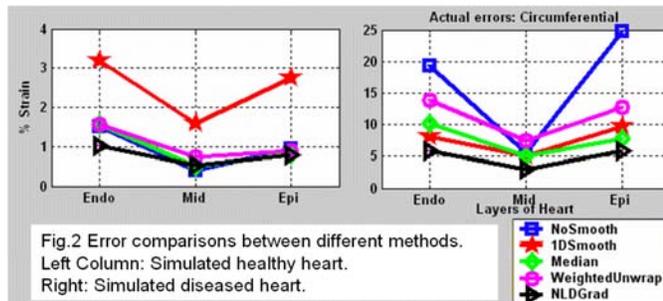
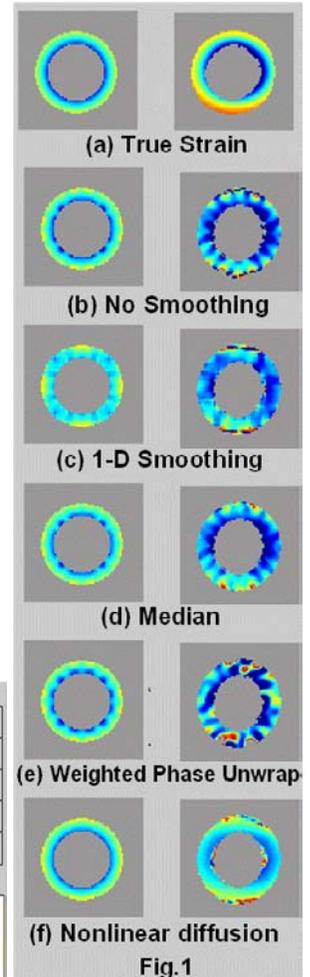


Fig.3: Examples of in-vivo E_{cc} strain mapping. Up: unsmoothed results. Bottom: Nonlinear diffusion smoothing. Artifact reduction is apparent after smoothing.