Artifact Reduction in HARP Stain Maps

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Introduction: The Harmonic phase (HARP) technique [1] was initially introduced as a computational approach for the automated analysis of tagged MR images. More recently, HARP MRI has developed into a fast imaging method for rapid imaging of cardiac function. HARP uses a bandpass filter to extract harmonic peaks in Fourier space. The phase of the resulting complex harmonic image contains information about motion. Past practice has shown that HARP strain images are corrupted by visible artifacts, often called zebra patterns, which degrade the quality of the strain measurements. Causes of this artifact include noise, Gibbs ringing, and interference from other Fourier peaks. The finite difference calculation of HARP is known to exhibit high variance [2] due to these sources, yielding artifacts in the computed strain maps. Osman and Prince [3] proposed to smooth the spatial derivatives of HARP using a 1D moving average filter weighted by the HARP magnitude images. The use of 1D averaging was justified by the need for the preservation of transmural changes in strain. But the size of the moving average filter is fixed, i.e. the size of the pass band of the corresponding low pass filter is fixed irrespective of the size of the bandpass filter used. Thus the 1D method does not consider the spectral characteristics of the strain. In this paper, we propose a 2D filtering approach which smoothes the unwrapped harmonic phase. The size of the 2 D filter will take into account the spectral characteristics of the strain.

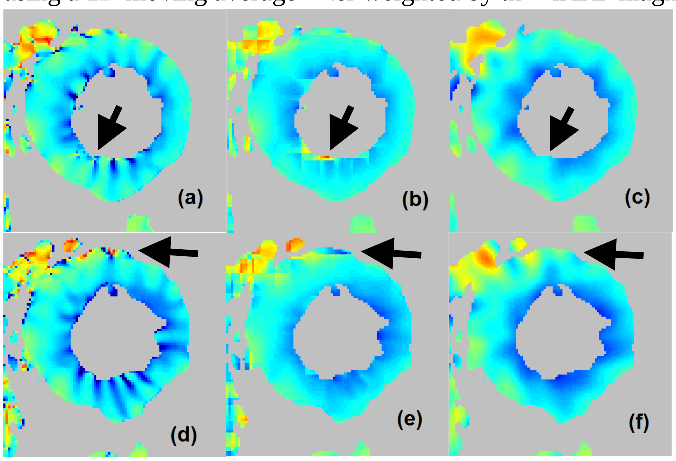

Fig. 1

Methods: The HARP images are first unwrapped using a weighted unwrapping algorithm [4], using a binary threshold of the HARP magnitude image as the weight. The use of this weighting image causes regions of high noise to be excluded in phase computation, which improves the estimate of the unwrapped phase. The resulting unwrapped phase is smoothed in the frequency domain using a low pass filter with the same size and shape as the harmonic bandpass filter. This removes higher frequencies created during the phase computation and unwrapping process but preserves the underlying strain resolution determined by the original bandpass filter [5]. The datasets shown in this abstract use FOV =280 mm and the size of the HARP bandpass filter is (40/FOV) mm^{-1}.

Results: Fig. 1 shows Eulerian strain maps calculated using three methods: traditional HARP method (Fig. 1(a) and (d)), the 1D moving average method (Fig 1(b) and (e)), and the proposed HARP unwrapping and smoothing method (Fig. 1(c) and (f)). The images are from a normal volunteer. The top and bottom row images correspond to 2 different mid systolic time frames in the cardiac cycle. Reduction of artifacts and the improvement over the 1D smoothing method are clear (see arrows). In this case, the size of the filter used in both the 1D filter case and the proposed method are the same. The region indicated by arrows shows the improvement over the 1D smoothing case. Because of the 2D smoothing, the blotchiness present in Figs. 1(b) and (e) are absent in Figs.1 (c) and (f). To show that the proposed smoothing method maintains the gradient in the radial direction, we plotted the Eulerian strain values in the endocardium, epicardium, and the midwall averaged over 8 octants for images in the second row of Fig.1. The results for the no smoothing, 1D smoothing and the unwrap-smoothing case are shown in Figs. 2 (a), (b), and (c), respectively. We can see that at each octant the strain gradient is maintained.

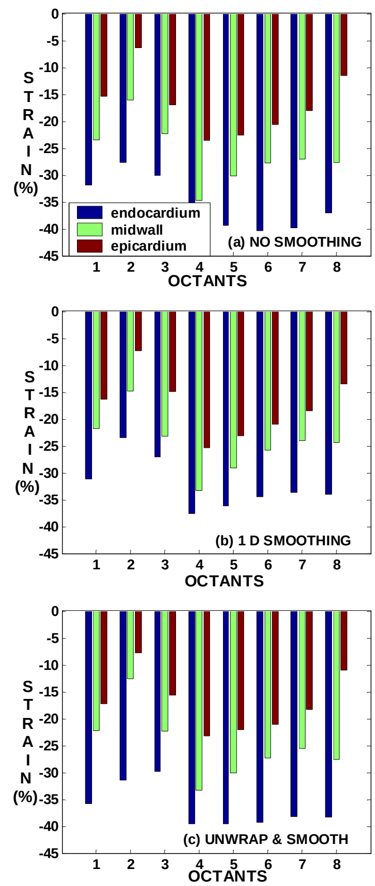

Fig.2

Conclusion: The presented method smoothes unwrapped HARP phase images. It is shown to reduce the artifacts in HARP strain maps while preserving transmural strain. The proposed method is better than a previously reported method.

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