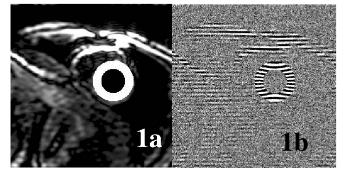
On optimal filter sizes for measuring cardiac motion using HARP-MRI

V. Parthasarathy[1] and J. Prince[2]

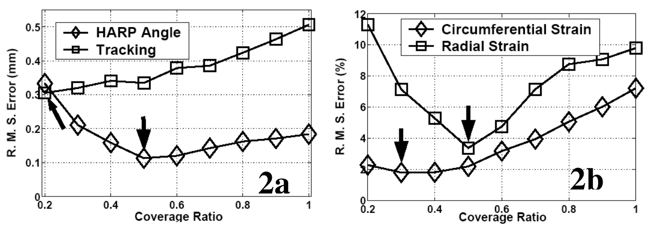
[1]Electrical and Computer Engineering, Johns Hopkins University, Baltimore, MD, United States, [2]Johns Hopkins University

INTRODUCTION: Harmonic phase (HARP) MRI uses band pass filters to extract harmonic peaks in the spectrum of the tagged MR image to automatically estimate myocardial motion and strain [1]. The size of the filter is critical to the accuracy of the motion estimates. If the filter is too small, then there is inability to measure large deformations and if the filter is too large, artifacts and noise overwhelm the measurements. Currently, in most studies that use HARP-MRI, the choice of filter is quite arbitrary – based entirely on visual inspection of the spread of the harmonic peak in the spectrum of the image. In this paper, we provide insights into how to select the filter size so that the overall error in the estimation of tissue motion is minimized. We simulate variety of imaging conditions by varying the amount of noise and strain levels in the myocardium. We optimize the filter sizes over these parameters. Our analysis is based on a computational cardiac motion simulator, which uses an analytical model of the heart's motion.

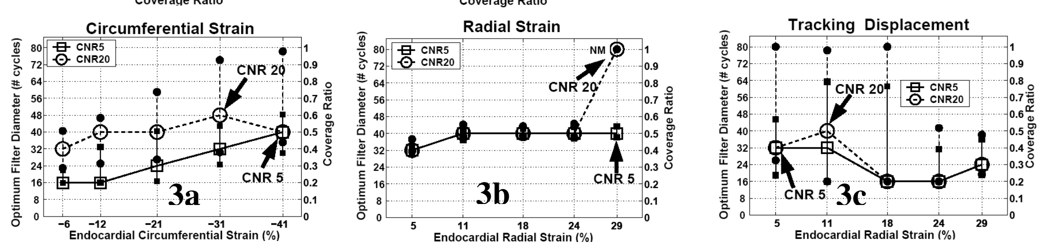


METHODS: Model: A computational cardiac motion simulator is used to simulate the motion of the left ventricle of the heart. The simulated contracting left ventricle is incompressible and is shaped like an annulus with inner radius R_i and outer radius R_o. During systole, the annulus deforms under radial contraction, where a layer with radius R shrinks to a smaller circle of radius $R_\varepsilon(R) = (R^2 + (2\varepsilon\text{-}\varepsilon^2) R_i^2)$, where the parameter ε controls the strain in the left ventricle. Fig.1(a) shows an example of short axis slice with a simulated left ventricle (see the annulus in the center). The background is simulated realistically using the harmonic magnitude image from an in vivo data set. This realistic background ensures that the spectral components of the simulated image are comparable to an in vivo image. In order to study HARP tracking, two consecutive time frames are simulated, such that first time frame has lesser strain than the second. Fig. 1a corresponds to the second time frame.

Experiments: CSPAMM protocol [2] was used to tag the magnitude images as shown in Fig. 1(b). A tag separation of 7 mm was used. Two different CNR values of 5 and 10 were simulated. For each CNR, 50 Monte Carlo realizations of random noise were performed and the results averaged. Five progressively increasing levels of strain were simulated to mimic the variation of strain end-diastole to end-systole. HARP-MRI was performed using nine bandpass filters of increasing sizes. The filter size is measured using the coverage ratio, which is defined as the ratio of radius of the filter and the tag frequency. The tag frequency is the number of sinusoidal cycles in the FOV that is FOV/tag separation. The nine filter sizes correspond to coverage ratios of 0.2, 0.3 …1. For each filter size, circumferential strain, radial strain, and tissue tracks were estimated and RMS error was calculated and averaged over the myocardium. The RMS errors plotted as a function of coverage ratio form cost functions for optimization.



RESULTS: Fig. 2 shows a set of representative optimization cost functions for CNR = 5 and mid-systolic strain ε = – 15%. The optimal coverage ratios are marked using arrows. The differences in the cost function for the different quantities are striking. Even though HARP angle is the basic measurement from which strains and tracking displacements are derived, the error in HARP angle is not directly predictive of the errors in strain and tracking. Therefore in the following, we optimize on the strains and tracking separately. A characteristic minimum is seen in Fig.2 (b) in the cost function of radial strain. When coverage ratios are small, the error is large due to loss of resolution. On the other hand, when the coverage ratios are close to 1, spectral interference and noise dominate the error. For tracking performance (Fig. 2a), we see the coverage ratios for optimal tracking are very small, unlike those of strains. This implies tracking can perform better with low resolution data than with high resolution data. Fig. 3 plots the optimal filter sizes for circumferential strain, radial strain, and tracking for two CNR values 5 and 10. The optimal filter diameters are plotted as functions of increasing strain during a cardiac cycle. The lengths of the tolerance bars (plotted on either side of the optimum) specify the ranges of filter sizes within which the errors will be minimum plus or minus 5% of the truth.



DISCUSSION: In all three cases, the optimal filter sizes for CNR 20 is greater than or equal to CNR 5. As CNR improves, the image noise decreases, and more of k-space can be included without losing precision due to noise. While the difference between CNRs is very distinct in circumferential strain, it is not distinct in radial strain and tracking. This implies that noise is major source of error in circumferential strains. The bars of radial strain, however, are much shorter than those of tracking, indicating a lack of robustness in radial strain estimates. Also, the tolerance bars for the circumferential strains and tracking, are long, which implies that both are quite robust to change in filter size. In terms of coverage ratios (right hand axis in Fig. 3), the optimum for radial and circumferential strain range between 0.4 and 0.7. The ratios are even smaller (0.3 – 0.5) in the case of tracking displacement. These ratios corresponds to optimum filter sizes of 40 plus or minus 10 samples in k-space, which explains the feasibility of acquisition protocols like FastHARP[2], which acquires only 32 x 32 samples around the harmonic peak.

CONCLUSION: The bandpass filters used in HARP should be selected carefully based on the following parameters: CNR, strain in the tissue, and type of measurement being done. Higher CNRs and higher strain values lead to larger optimal filters. Radial strain is sensitive to errors, both from loss of resolution due to small filters and from the loss of precision due to large filters. Therefore a coverage ratio of 0.5 is ideal for estimation of radial strain. On the other hand, tracking and the estimation of circumferential strain is very robust to change in filter sizes and can be estimated optimally using smaller filters. **REFERENCES:** 1) Osman et al, Phys. Med Biol, '00, 2) Fischer et al, MRM 1993 3) Sampath et al, MRM 2003