HARP-MRI Detectability — Contrast Detail Analysis using Simulations

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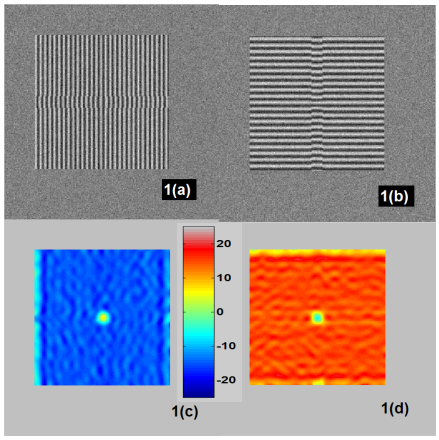
Introduction: Harmonic phase (HARP) MRI uses bandpass filters to extract harmonic peaks in the spectrum of the tagged MR image to automatically estimate myocardial motion and strain [1]. In previous work, an edge model in strain was used to arrive at the conclusion that the resolution of the HARP strain maps is approximately equal to the intrinsic Fourier resolution defined by the bandpass filter [2]. This edge model, however, does not address the *detectability* of abnormal tissue strain in small regions. In this abstract, we use contrast-detail analysis on simulated tagged MR images in order to study strain detectability in HARP. In particular, we use computer simulations to address the question of what is the smallest abnormal tissue that be detected for a given true strain contrast between healthy and abnormal tissue.

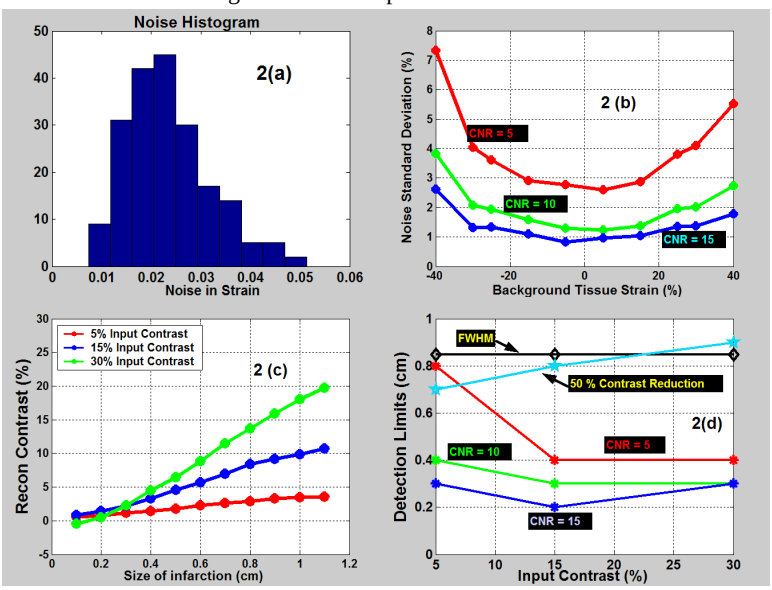


Methods: ***Simulation:*** We simulated a localized change in strain by placing a (simulated) tissue with no compressibility in the center of a healthy tissue, simulating an infarction or necrosis. The resulting tissue was tagged with vertical and then horizontal tags, as shown in Figs. 1a and 1b. The images were simulated using a FOV = 28 cm, resolution = 1 mm, and tag spacing = 0.7 cm. We assume CSPAMM acquisition with flip angle compensation so that tag contrast remains constant. A HARP bandpass filter of size 33 x 33 was used. The tissue was compressed in the horizontal direction and expanded in the vertical direction, and the tissue in the center had zero strain. Horizontal and vertical components of strain, computed using HARP were then computed, as shown in Figs. 1c and 1d, respectively. In order to perform a contrast-detail analysis, the size of the infarction (the detail) was varied from 1.1 cm to 0.1 cm with varying background strain (contrast) of –5%, –15%, and –30% in the horizontal direction. The size of the background was 8 cm on either side of the abnormality.

Noise Experiment*:* The detectability of abnormal tissue depends on the amount of noise in the background of these strain maps. Noise in HARP strain maps comes from two sources: 1) systematic sources like interference from the neighboring spectral peak and Gibbs ringing (which we model as noise) and 2) random image noise. In order to quantify noise arising from these two sources, we compute the same simulation without the abnormal tissue, thereby simulating a healthy tissue in a remote region of the heart. We then perform 200 Monte Carlo trials of the imaging process with added Gaussian noise such that CNR = 5, 10, and 15. In each trial, a 1 cm x 1 cm region was analyzed for the deviation from the true strain and the histogram of this deviation is shown in Fig. 2a. The offset of the histogram quantifies the bias due to systematic errors and the standard deviation of this histogram shows the error due to image noise. For use in the contrast detail analysis, the following noise threshold was used: n = bias + 2*(standard deviation). We say that the abnormal tissue is detectable if the detected contrast is greater than the noise threshold, n.

Results: ***Noise Results:*** Fig. 2b shows a plot of the noise thresholds as a function of background strain for different CNRs. Because these errors are based on the interference with the neighboring peak, the error changes with background strain. We see the errors are in the range of 1–7 % and they increase with decrease in CNR and increase in background strain. This is explained by the fact that as the strain increases, more 'motion signal' goes outside the HARP bandpass filter and image noise and spectral interference become more prominent.

Contrast Results: Fig 2c shows the percent contrast as a function on of infarction size for three different true strain contrasts. It is observed that larger true strains and larger infarction sizes are more detectable. Fig. 2d shows the minimum size that an infarct must be, in order to be detectable. Three different CNR's are shown and the FWHM (black line) is plotted for comparison to the intrinsic resolution. Also plotted (in cyan)is the size of the infarction at which the contrast reduces to 50% of the truth. It is observed that infarcts can generally be detected, even if they are much smaller than the FWHM and the 50% limits. For CNR = 15, a common value given for CSPAMM tagged MR images, infarcts as small as 0.3-0.5 times the FWHM can be detected.



Discussion and Conclusion: This simulation study suggests that regions of abnormal strain that are considerably smaller than the FWHM of the imaging system can be detected using HARP. This conclusion must be explored further, however, since other factors such as object geometry (including background size) and the direction of strain computation as well as other factors are not modeled and are likely to yield different results. Further simulations and phantoms studies will shed further light on the issue of detectability in HARP.

References:1) Osman et. al, *Phys. Med Biol*, 45:1665-1682, '00 2) Parthasarathy ,et. al abstract *ISMRM*,'04