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Automated Correction of Background Intensity Variation and Image Scale Standardization in 4D Cardiac SPAMM-MRI

A. Montillo[1], L. Axel[2], D. Metaxas[3]

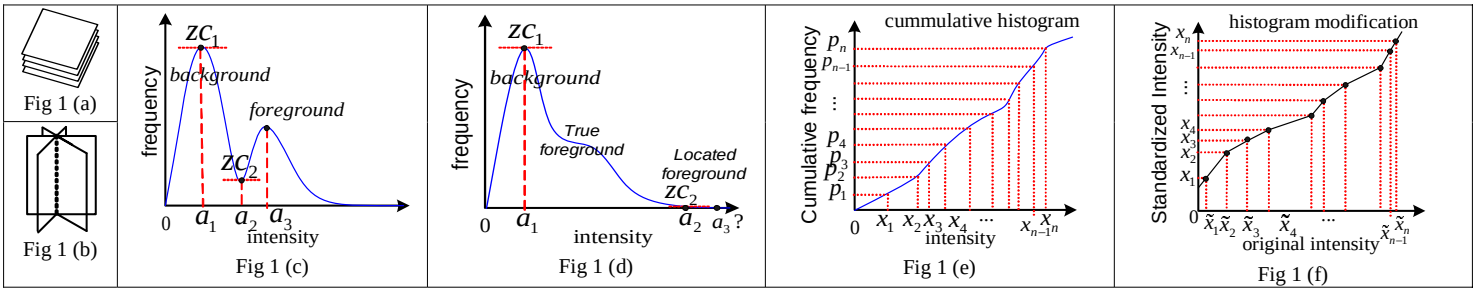
[1]University of Pennsylvania, Philadelphia, United States, [2]New York University, Radiology Dept, NY, NY, United States, [3]Rutgers University, Piscataway, NJ, United States

Introduction

Cardiovascular disease is the leading cause of death in the western world. However SPAMM-MRI, an imaging method that provides an in-vivo measurement of regional heterogeneity of myocardial contraction, can facilitate patient diagnosis and treatment planning. Widespread adoption of SPAMM-MRI is hindered by the 5+ hours required to analyze ~300 images per subject. This abstract presents methods that dramatically reduce SPAMM-MRI's salient artifacts: (1) severe intensity inhomogeneity from surface coils and (2) tissue intensity variation (int***er***subject and int***ra***subject during contraction). These methods open the door to fully automated algorithms for SPAMM analysis. Results are presented for normal subjects and ventricular hypertrophy.

Methods

Images were acquired using a 1.5T GE cardiac MR scanner. ECG-gated gradient echo imaging was used, with 30ms between imaging phases (times). At each phase, 12 parallel short axis (SA) images were acquired aligned to an axis through the center of the left ventricle and 9 long axis (LA) images were acquired along planes with an angular separation of 20° and whose intersection approximates the long axis of the left ventricle. It has been shown [1] that intensity inhomogeneity can be effectively suppressed in SA volumes (Fig 1a) using a method that determines the set of voxels that are most homogeneous based on intensity and intensity gradient measures. In [1] a 2nd order polynomial is fit to the intensities in this set, and the image is corrected by dividing the entire image by this fitted inhomogeneity function [2]. The process is iterated until there is no change in the set of voxels. This method assumes the voxels are aligned to a Cartesian coordinate system. *To adapt this method* for severe inhomogeneity of LA volumes (Fig 1b) whose pixels are not Cartesian aligned, we fit the polynomial to each slice individually. In addition we scale the 8 bit images to 16 bit images: $I_{16} = (I_8 - \min_{x,y} I_8)/(\max_{x,y} I_8 - \min_{x,y} I_8)4095$ to retain the precision in subsequent iterations which involve dividing by the polynomial.



Histogram modification has been used to standardize the intensities of MR images of the human brain [3]. The images from a scan of a subject, with well delineated tissues, is chosen as the standard scan and key landmarks are located on the intensity histogram from that subject. To standardize intensities on subsequent images from the same subject or from new subjects, the same landmarks are located in the histogram of the new scan and linear interpolation is used to define a mapping from the original intensities in the new scan to output intensities. This makes the subsequent scan have roughly the same histogram as the standard subject. Essential to this algorithm are the histogram landmarks and any assumptions about the shape of the histogram. The method in [3] assumes that the histogram is *bimodal* (Fig 1c) and that the second zero crossing, zc_2, of the slope of the histogram can be used to locate distribution of the foreground pixels. In SPAMM-MRI, blurring and fading of tag lines during systole quickly fills in this histogram valley yielding a *unimodal* histogram (Fig 1d). To handle both bimodal and unimodal histograms seamlessly, we compute landmarks based on percentiles $(x_1,\ldots,x_n)$ in the *cumulative* histogram of the standard subject scan (Fig 1e) and the percentiles $(\tilde{x}_1,\ldots,\tilde{x}_n)$ in any new scans. 16 percentiles were empirically determined to be sufficient. Using linear interpolation, a mapping function is defined to transform original intensities from a new scan to standardized intensities (Fig 1f).

Results

We first apply the intensity *inhomogeneity correction* algorithm to the images obtained from the scanner. This removes the slowly varying intensity due to the field inhomogeneity from the surface coils. Then we select the histogram from one phase of one subject to be the standard and apply *intensity standardization* to transform the histogram of the other subjects to match the standard. To validate, we compare the images for each subject, before and after the application of the algorithms. Fig 2a shows the significant effect of the correction algorithm: top row contains LA images before correction for a subject with ventricular hypertrophy, bottom row are the intensity flattened images after correction. Fig 2b demonstrates the int***ra***subject standardization that occurs over time since we standardize all phases to one particular phase of the standard subject. The top row shows one image from each volume taken over time (from left to right, phase=2,3,4) after standardization using zc_2 to locate the foreground. Phase 3 and 4 images are too dark because histogram is unimodal. When we use the new percentile method, described above, all phases are properly standardized (bottom row). Fig 2c demonstrates int***er***subject standardization, top row contains images after correction but before standardization for 3 different subjects. Bottom row: after correction and standardizing all subjects, the myocardiums have similar intensity. Similar results are found in the rest of our 20 subject database.

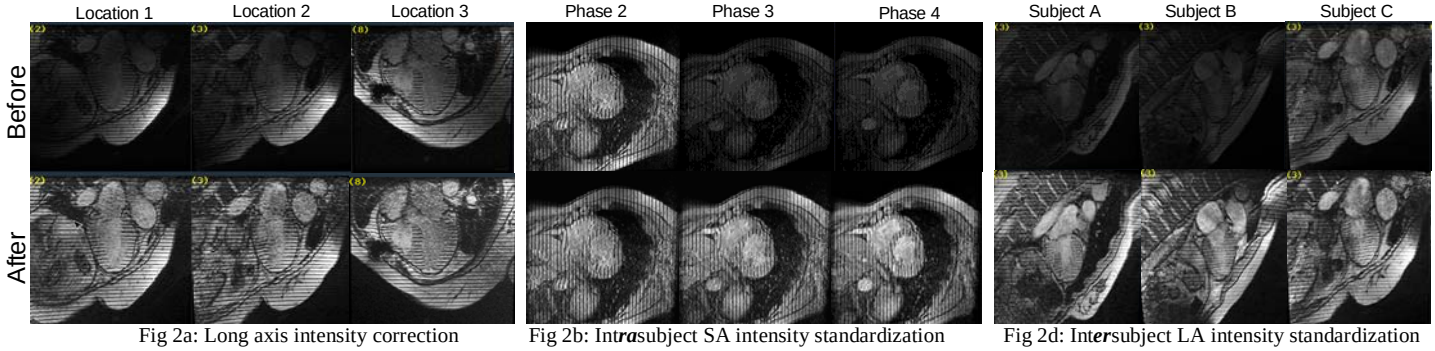


Fig 2a: Long axis intensity correction

Fig 2b: Int***ra***subject SA intensity standardization

Fig 2d: Int***er***subject LA intensity standardization

Discussion We have presented a fully automatic method to correct intensity inhomogeneity in long axis cardiac MR images. The quality of the resulting images is significantly superior to the original images for use with an automated segmentation algorithm or manual contour tracing. We have presented a novel method to standardize the intensities of tagged cardiac MR images which have both uni- and bimodal intensity histograms. Both intersubject and intrasubject tissue intensity deviations are significantly reduced after correction in SA and LA images. We have found good results for both normal and diseased subjects. Future work includes extending the correction method to estimate a 3D inhomogeneity field using a cylindrical coordinate system, and extension to incorporate wraparound artifacts.

References

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