ISSN: 2278 – 909X
International Journal of Advanced Research in Electronics and Communication Engineering (IJARECE)
Volume4, Issue 6, June 2015

STUDY THE ENERGY MINIMIZATION TECHNIQUES IN ORDER TO IMPROVE THE QUALITY OF AN MIR IMAGE

Anamika
Electronics & Communication Engg. Department
Maharishi Markandeshwar Engineering College,
Mullana(Haryana), India

Dr. Rajat Gupta Dr. Kuldip Pahwa
Electronics & Communication Engg. Department
Maharishi Makandeshwar Engineering College,
Mullana(Haryana), India

Abstract: This Paper proposed the characterization of atherosclerosis lesions in vivo using Magnetic Resonance Imaging (MRI). For this, it uses an energy minimization approach to equalize the contrast of MR image. To enable the use of voxel gray values for interpretation of disease, we created a new method, energy minimization with a spline model, to correct the severe (80%) intensity in homogeneity that arises from the surface coil array. Intensity in intravascular micro coils induces shading artifacts across the data. Artifacts are modelled as a smooth multiplicative bias field with a cubic spline. The cubic spline is optimized to minimize the entropy of the correct image The entropy based method does not require classification and robustly addresses some problems that are more severe than those found in brain imaging, including noise, steep bias field, sensitivity of artery wall voxels to edge articrafts, and signal voids near the artery wall. All simulations are done in MATLAB.

Keywords – Mr Images, energy minimization, blood vessels, optimization etc.

I. INTRODUCTION

Cardiovascular disease (CVD) us the leading cause death and disability in the western world, accounting for 40% of the nation's mortality and much of its morbidity [1]. The American Heart Association estimates that 7.13 million Americans had CVD in 2003. Approximately 1 in 3 initial heart attacks is fatal, and in 1 every 6 patients who experience a heart attack, sudden death is the first, last and only symptom.

Atherosclerosis is an inflammation of the intima layer of the arteries that induces thickening of the vessel walls through the development of a plaque. It is ubiquitous, and starts early in life almost half of the infants in their first 8 arteries as well as twice the number of macrophages without lipid droplets, and among children aged 2 to 15 years, 99% have type II lesions (fatty streaks) in their aortas. Atherosclerosis occurs principally in large and medium sized elastic and muscular arteries, can lead to ischemia of the heart, brain, or extremities, and may induce infarction. Atherosclerosis may be present throughout a person's lifetime. [1]

Normally, cholesterol transits in the blood stream packaged in droplets which are internalized or secreted by cells to maintain lipid homeostasis; VLDL, LDL, HLD, for very low, low, and high density lipoproteins respectively. Lipids are ubiquitous molecules present in all cells. After their assimilation in the intestine they are transported to the liver in the form of chylo-microns. Lipid intake from food represents about 10% of the daily need. The remaining part is produced by the liver. Lipids can then be transported from the liver to other tissues in the LDL and VLDL forms, through the Reverse Cholesterol Transport which is the only way for peripheral tissues to get rid of cholesterol. [2]

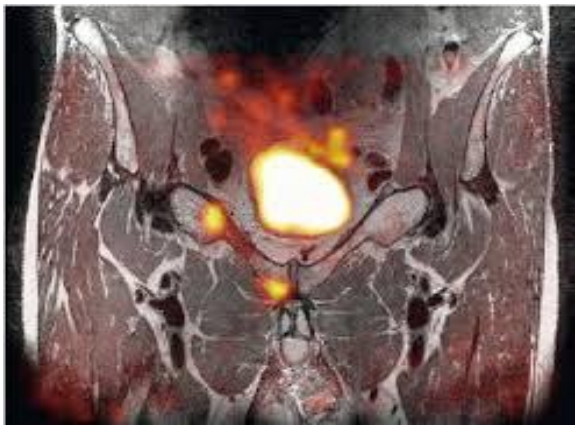


Figure 1: MR Image [2]

The trigger of atherosclerosis is the infiltration of LDL in the intima. This can result solely from very high levels of LDL, from endothelial dysfunction caused by the factors listed above, and is often found at bifurcations probably because of non-laminar flow and/or increase shear stress. Endothelium injury leads also to compensatory responses that after the normal homeostatic properties of the endothelium. The Different forms of injury increase the adhesiveness of the endothelium with respect to leukocytes or platelets, as well as its permeability. The injury also anticoagulant properties and to release vasoactive molecules, cytokines, and growth factors. [3]

The paper is organized as follows. In Section II, It describes MR images. In section III, it describes the proposed steps used in the processing technique. The results are given in Section IV. Finally, conclusion is explained in Section VI.

II. MR IMAGES

MRI is a priceless procedure that utilizes radio waves, powerful magnet and a computer to detect detail images, Our body is made up of millions of hydrogen atoms (i.e. 80% of water) which are magnetic in nature. When our body is placed in magnetic in nature. When our body is place in magnetic filed these atoms align in the field, much like a compass point to the North pole. A radio wave "knocks down" the atom and disrupts their property. [4]

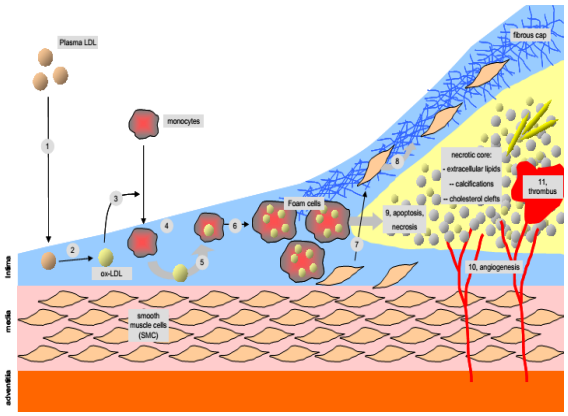


Figure 2: Major events of atherosclerosis progression [4]

Magnetic resonance imaging is based on many physical properties of tissues and can achieve exquisite contrast of soft tissues, without any radiation and in a completely non invasive way. The most common imaging methods that have been used to image atherosclerosis and stenosis are based on spin-lattice relaxation (T1), spin-spin relaxation (T2), proton density (PD), diffusion, magnetization transfer (MT), flow (time-of-light): TOF and phase contrast : PC) and gradient echo (GRE)[5]

One of the first goals of arthrosclerosis imaging has been for a long time to identify and quantify stenosis of vessels. When a plaque develops, the lumen may be obstructed thereby reducing the blood flow to the downstream regions, resulting in ischemia. However, because the disease is chronic and plaques develop over years. Secondary circulation is often recruited and day-to-day life can be asymptomatic. [6]

Three main imaging modalities exist to quantify stonesis: Ultrasound imaging, X ray imaging, and MRI. Early X-ray systems used a contrast agent and planar projections to generate different views which, after interpretation by a trained expert, produced an estimation of the extent of lumen narrowing. More recent techniques use 3D CT (CTA), but stil require the injection of a contrast agent opaque to X-ray. One advantage of CTA is the clear visualization of calcifications and ossifications, almost always associated to advanced atherosclerotic lesions [16]. The major drawback of X-ray modality is the use of ionizing radiations. This problem can be avoided by using magnetic resonance angiography (MRA). MRI is safe and allows imaging of the vasculature either with a contrast agent, or in absence of a contrast agent owing to the following properties of the blood. [7]

MRI presents the advantages of allowing simultaneous visualization of the arterial lumen and vessel wall, and further has the potential to characterize atherosclerotic plaques. Major plaque components, such as the lipid core, calcification, fibrous connective tissue, and intra-plaque hemorrhage/thrombus have been identified with regard to their signal intensity characteristics on black blood (where blood signal is canceled and the lumen appears thus black) T1W, T2W and PDW images. These black blood MRI techniques are also useful for vessel wall area measurement, a generally-accepted direct measure of plaque burden.[8]

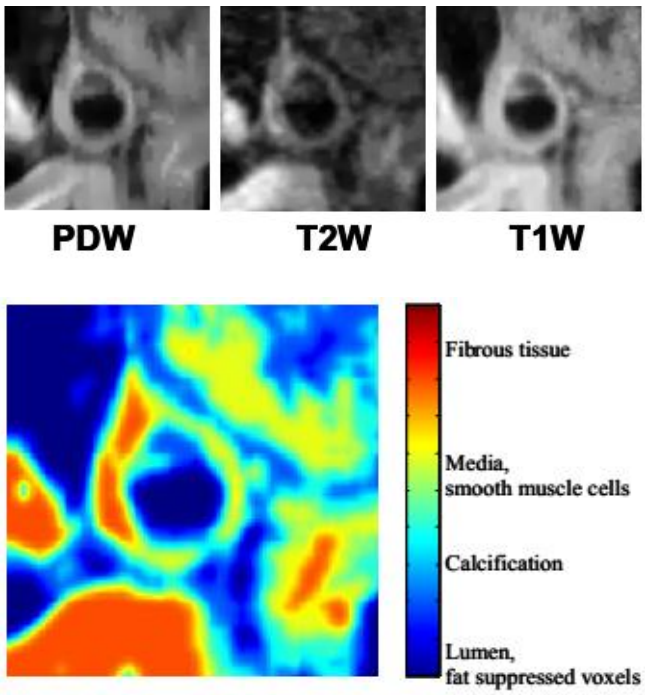


Figure 3: Tissue Classification [9]

Two major contrast agents are used in MRI: Gadolinium-based molecules that decrease T1 relaxation times, and small iron oxide particles that disturb the magnetic field. The former has a diameter of about 0.8 nm for the common Gd-DTPA. The latter has many forms that span diameters from 4nm to 1000 nm or bigger. SPIO have a typical size of 70-140 nm. For comparison Albumin has a diameter of about 7 nm and a size of 68 kilo Dalton. Gd-DTPA contrast agents are small molecules that can diffuse through the endothelium over time, whereas the bigger iron oxide particles require a much longer time to do so. In order to get a significant sensitivity with MRI with a CA, a concentration of 1 mm is necessary. This concentration is obtained after 1 hour when particle size is about 10nm. Therefore, these MRI contrast agents will not penetrate cells nor diffuse significantly through the normal endothelium barrier in the time compatible with clinical imaging. [10]

So, from above details, multi-contrast MRI can characterize atherosclerotic lesions in vivo. Beyond expert examination of multiple images taken at the same location, whether they are regular spin-echo acquisitions, more advanced sequences like magnetization transfer, or contrast-enhanced data set using recent molecular imaging advances, computer methods are promising to reduce substantially the time and therefore the cost of analyzing such big sets by providing the physicians with a reliable and accurate meta information display of the underlying disease [10]

III. IMPROVE QUALITY OF MR IMAGES BY ENERGY OPTIMIZATION

A critical step for analysis methods relying on voxel gray values is the correction of the signal intensities across the MR image. The principal source of the degradation is the spatial in-homogeneity of coil sensitivity of the specially designed surface coils. A correction algorithm for carotid artery imaging faces many challenges. First, the receive coils suffer from a very steep sensitivity fall-off in the direction of increasing tissue depth that is much more significant than the variation across the brain when imaging with a head coil. If not well corrected, this can confound the examination of the vessel wall by experts and defeat automatic tissue classification algorithms. Second, the noise present in the MRI carotid images can disrupt algorithms. Third, there are many voxels close to the artery walls that are void of signal, wither from fat suppression or from blood flow compensation.[11]

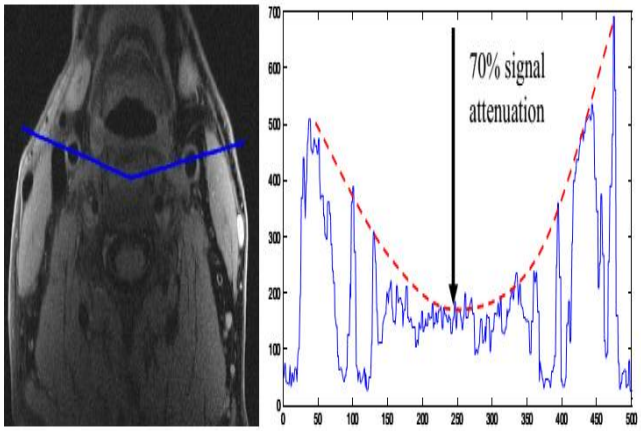
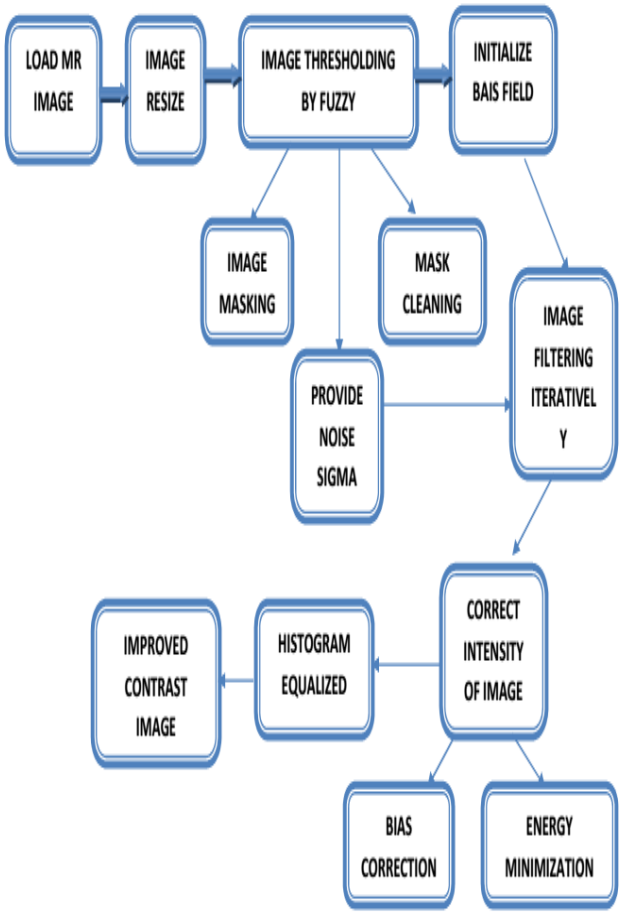


Figure 4: Example of intensity inhomogeneity [11]

MRI parameters, like many other imaging modalities, must be chosen with the trade-off between noise in resolution always present. Vessel wall and plaque features are small, on the order of few millimeters, requiring sub-millimeter resolution in the carotid arteries, and probably closer to 0.1 mm for imaging coronaries. MRI resolution can be decreased but at the expense of time or noise. Limited resolution has two consequences. First, sharp edges between two distinct tissues are blurred (e.g. lumen/vessel interface). Second, anatomical structures smaller than the voxel size disappear (e.g. small calcification inside the lipid core). This problem is exacerbated by the noise: when a voxel is slightly brighter than its surrounding voxels. We tried a new approach to this problem based on non-linear diffusion filtering than can reduce partial volume effect when it is possible, without generating artifacts or spurious features when it is not. We have also extended the technique by incorporating a noise reduction term.[12]

We explored whether sub-voxel registration, either rigid or non rigid, could be achieved using these techniques and discovered that in most cases the answer was negative. That is, common similarity measures that are minimized to compute the best registration parameters suffer from artifacts that result in local minimum and a global bias. We explored a new accurate method to validate MRI methods using energy optimization. We developed and validated a more accurate method based on the novel 3D imaging device developed.[12]

Steps of Proposed System:



We first identify all tissue voxels and filter the image to reduce noise. We fit a third order polynomial function to the tissue voxels so as to provide a rough initial estimate of the bias field, B0. Air voxels in the background are excluded because they are void of signal. For the refined description,

we model the bias field, B, as a spline with a recetangular grid of knots evenly spaced across the image.

We now describe the piecewise optimization process which makes the use of a bicubic spline model tractable. We identify the knot k1 having the highest corresponding B0 value and begin optimization there. The signal from the coil we will get a good local estimate of B.

IV. RESULTS

We first identify all tissue voxels and filter the image to reduce noise. We fit a fourth-order polynomial function to the tissue voxels so as to provide a rough initial estimate of the biasfield. Air voxels in the background are excluded because they are void of signal. For the refined description, we model the bias field, as a bicubic spline with a rectangular grid of knots evenly spaced across the image. The spacing of knots is important: knots should be sufficiently close to ensure that the bias field can be adequately expressed and far enough apart that the estimated surface will not contain anatomical structures in the images. Spacing is related to the receiver coil geometry and is optimized in experiments described later. We use the values of the initial polynomial estimate of the biasfield at the knot locations to initialize the bicubic spline biasfield.
We now describe the piecewise optimization process which makes the use of a bicubic spline model tractable. We identify the knot having the highest corresponding value and begin optimization there. The signal from the coil at this location is high and the high local SNR ensures that we will get a good local estimate of B.

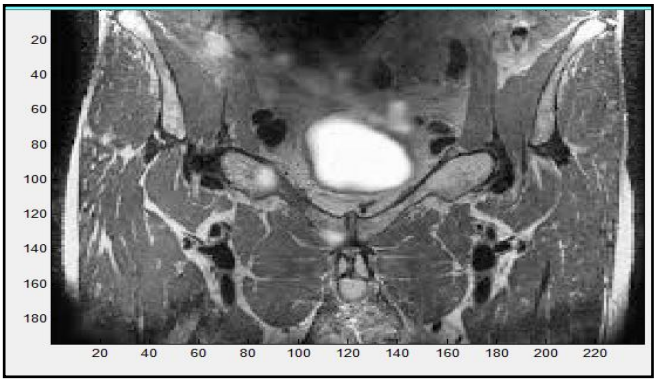


Figure :) Input image

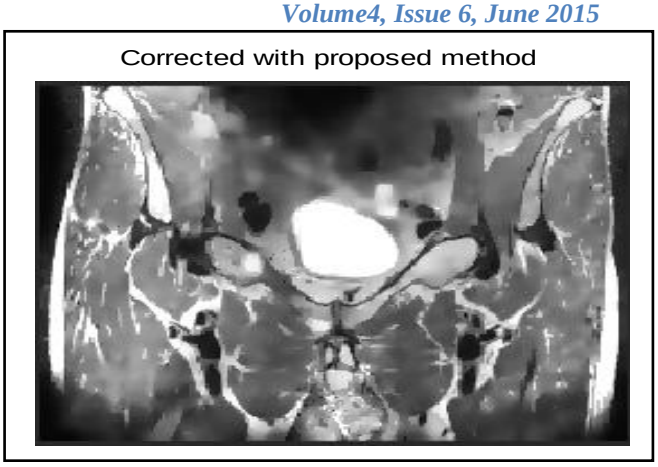


Figure :) Output image

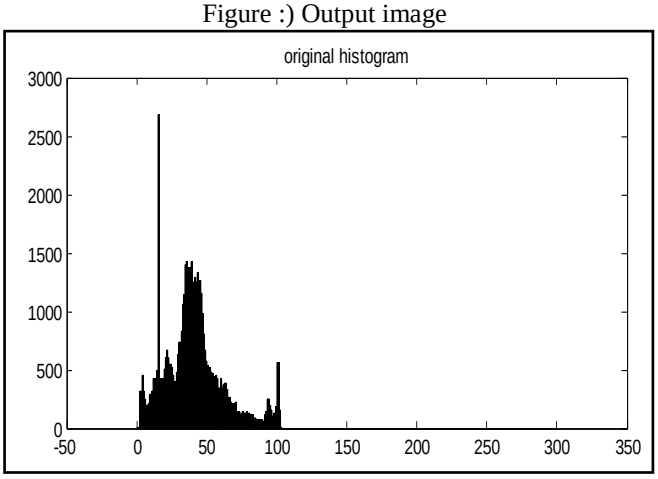


Figure :) Histogram of original image

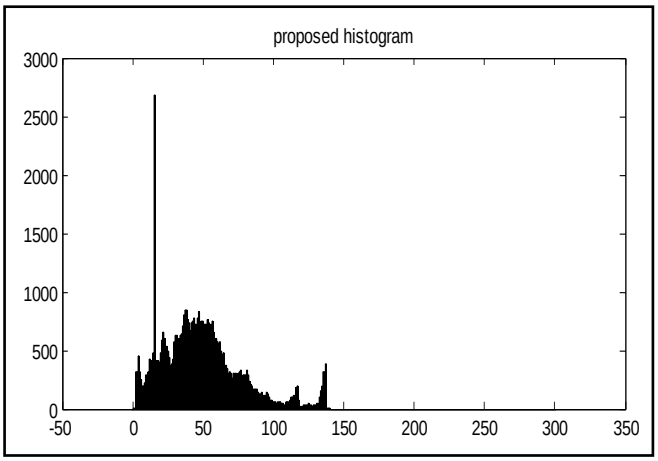


Figure :) Histogram of output image

No. of Voxels	Mean	S.D.	Threshold
4	42.03	7.60	64.85
3	42.81	6.58	62.36
2	42.85	6.22	61.22
1	42.71	5.35	58.76
0	42.66	5.36	58.75
Other Performance Parameters			
Entropy		3.9521	
Change on Knots		0.4522	
Change on Entropy		-0.517	

Performance parameters of system

V CONCLUSIONS

In this work, it presented a variational level set framework for segmentation and bias correction of images with intensity in homogeneities. Here, described a new method called energy minimization to correct for intensity inhomogenites of images. By optimizing a cubic spline to minimize the entropy of a dataset, a bias field can be estimated and the images corrected. This proposed method is designed for those difficult cases and showed good results for surface coils, but also for interventional MRI, probably the most extreme case of inomogeneity.

REFERENCES

[1] R. Gonzalez, R. Woods and S. Eddins, “Digital Image Processing Using Matlab”, 2004, Prentice Hall.
[2] Gonzalez, R.C., Woods, R., “Digital Image Processing”, 2nd Edition, Prentice-Hall, (2002).
[3] Archana R P, “Medical Application of Image Segmentation with Intensity Inhomogeneities” IJSER 2014.
[4] Wenbin Li, Darren Cosker, Matthew Brown, Rui Tang, “Optical Flow Estimation using Laplacian Mesh Energy”, IEEE 2013.
[5] Xiaodi Hou, Jonathan Harel, “Image Signature: Highlighting Sparse Salient Regions”, IEEE 2012.
[6] Yuanjie Zheng, Stephen Lin, Sing Bing Kang, “Single-Image Vignetting Correction from Gradient Distribution Symmetries”, IEEE 2012.
[7] Chunming Li, Rui Huang, Zhaohua Ding, J. Chris Gatenby, “A Level Set Method for Image Segmentation in the Presence of Intensity Inhomogeneities with Application to MRI”, IEEE 2011.
[8] Zachary T. Harmany, Roummel F. Marcia, “This is SPIRAL-TAP: Sparse Poisson Intensity Reconstruction Algorithms – Theory and Practice”, IEEE 2010.
[9] Mila Nikolova, Michael K. Ng, and Chi-Pan Tam, “Fast Nonconvex Nonsmooth Minimization Methods for Image Restoration and Reconstruction”, IEEE 2010.
[10] Jiejie Zhu, Liang Wang, “Spatial-Temporal Fusion for High Accuracy Depth Maps Using Dynamic MRFs”, IEEE 2010.
[11] Stamatios Lefkimmiatis, Petros Maragos, “Bayesian Inference on Multiscale Models for Poisson Intensity Estimation: Applications to Photon-Limited Image Denoising”, IEEE 2009.
[12] Gheorghe Postelnicu, Lilla Zöllei, and Bruce Fischl, “Combined Volumetric and Surface Registration”, IEEE 2009.