PLATELETS FOR DENOISING OF SPECT IMAGES: PHANTOM AND PATIENT STUDY

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Abstract: In this study the evaluation of a Platelet-based Maximum Penalized Likelihood Estimation (MPLE) for denoising SPECT images was performed and compared with other denoising methods such as Wavelets or Butterworth filteration. Platelet-based MPLE factorization as a multiscale decomposition approach has been already proposed for better edges and surfaces representation due to Poisson noise and inherent smoothness of this kind of images.
We applied this approach on both simulated and real SPECT images.For Nema phantom images, the measured noise levels before (2.1732) and after denoising with Platelet-based MPLE approach (0.1399) was found.In patient study for 32 cardiac SPECT images, the difference between noise level and SNR before and after applying approach were (M_b=3.7607, SNR_b=9.7762, M_a=0.7374, SNR_a=41.0848). Thus the Coefficient Variance (C.V) of SNR values for denoised images with this algorithm incompare with filteration with Butterwoth filter, 145/33% was found.For 32 brain SPECT imagesvalues this Coefficient Variance,196/17% was found.
Our results showed that Platelet-based MPLE is a useful method for denoising SPECT images considering better homogenous image,improvements in SNR, better radioactive uptake in target organ and reduction of interfering activity from background radiation in compare with that of other conventional denoising methods.
Keywords:Denoising, Image Approximation, SPECT, Platelets, MPLE

Introduction

In photon-limited applications such as PET, SPECT, Infrared (IR) imaging, astronomical imaging, the basic problem is low S/N ratio compared to other imaging systems because of small number of detected photons. The sources of noise are low count levels scatter, attenuation and electronic noises in the detector/camera[1]. Maximum Likelihood Estimator (MLE) is the most popular tool for photon-limited image denoising but image appears noisy and has a large variance about its true mean. also filtering of the MLE reduces the noise at the expense of blurring important details in image(edges in the image)[2].

Robert D. Nowak and Rebecca M. Willett proposed a multiscale approach for recovering edges and surfaces in photon limited medical imaging [3].

Platelet-based maximum penalized likelihood criterion has been introduced for image denoising and deblurring in this field of imaging systems. a review of this method can be seen in appendix.

Materials and Methods

A. Phantom Study

We applied the platelet-based maximum penalized algorithm on two kinds of Phantoms and compared the results with wavelet and Butterworth filtration.

A-1. Monte Carlo Simulated Phantoms

All Monte Carlo simulation were generated with the SIMSET package to model the physical processes and instrumentation used in emission imaging [4],[5].
we made hot and cold bar phantoms in both low noise state(high activity and big time of imaging)and high noise state(low activity and small time of imaging).

For cold phantoms the target cylinder is 8cm long with a radius of 20cm that is filled with activated water (usually 20mCi of ^{99m}Tc). inside, there is a cylinder with thickness of 8 .5 cm was all made of iron to indicate cold defect .

For hot phantoms the target cylinder is 8cm long with a radius of 20cm that is filled with water. There is a cylinder with radius of 8 .5 cm made of iron and filled by activated water to indicate hot defect.
figure(1)shows a typical high noise hot simulated phantom.

All the phantoms were reconstructed by Filtered Back projection (FBP) algorithm using MATLB software.The evaluation was performed quantitatively on phantom data with Power spectrum and noise level measurements between reconstructed

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image and denoised image with Platelet-based MPLE, wavelet and butterworth filteration. Estimating the noise level can be done by Median Absolute Deviation method using:

$$\hat{\sigma} = \frac{1}{0.6745} MAD\ (\{d_{j,k}, 0 \le k \le 2^{j}\}) \qquad (1)$$

In Eq(1), MAD is Mean Absolute Deviation and dj,k is the wavelet coefficients of the finest level .

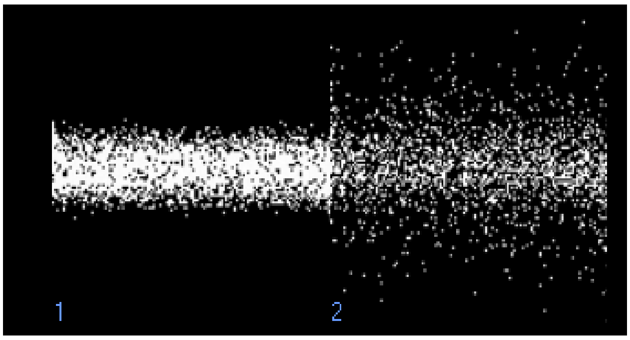
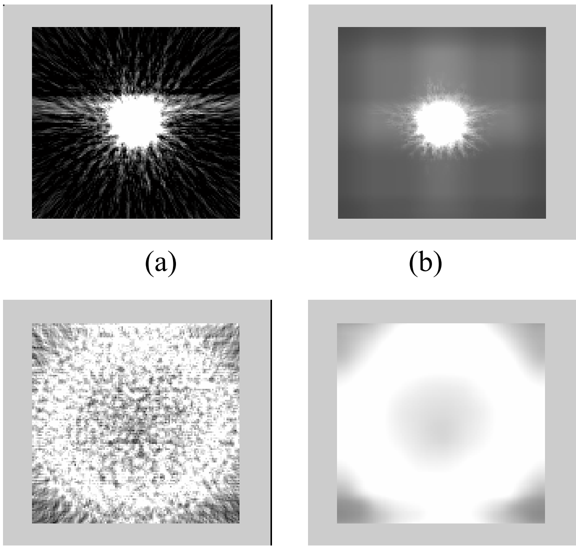


Figure 1: Image of a hot simulated phantom:
1) Without scattering image, 2)Scattering image



(a) (b)

(c) (d)

Figure 2: (a) Reconstructed image of hot phantom, (b)Denoised image of (a) with Platelets, (c)Reconstructed image of cold phantom, (d) Denoised image of (b) with Platelets.

A-2. Real Phantom

We used the NEMA phantom which meets the specifications for measuring the system's Line Spread Function according to NEMA protocol. Phantom consists of a cylindrical tank, (20.3 x 20.3 cm)diameter, made of Acrylic

with three stainless steel tubes with 200mm hight,200±5mm diameter and 2mm±10 wall thickness. They were filled by 2 mCi of ^{99m}Tc in such as amount of counts were less than 20000 counts per second.

The tank was filled with water to provide proper gamma-ray scattering and was suited in the way that the longitude axis of phantom was matched with the longitude axis of the bed of SPECT system and the longitude axis of detector rotation.

Imaging was performed using a single-head, Argus model SPECT system . Images were recorded over 360˚ in 128*128 matrices with an acquisition time of 5 seconds per projection. 32 SPECT images were obtained.

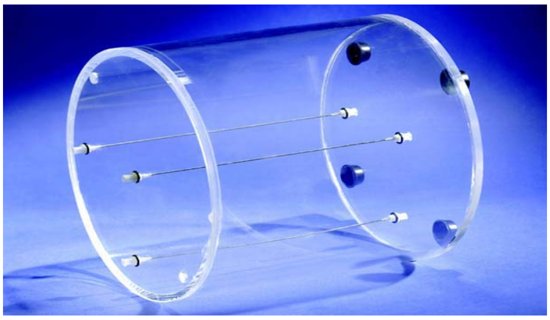


Figure 3 : NEMA Phantom

B. Patient Study

32 cardiac SPECT images were obtained using a dual-head, ADAC model SPECT system .A low energy general purpose collimator was used for this purpose. Images were recorded over 180˚ from 45˚ right anterior oblique to 45˚ left posterior oblique in 64*64 matrices with an acquisition time of 25 seconds per projection . In clinical practice about 20 mCi of ^{99m}Tc is injected into patient.

In another study 32 brain SPECT images were obtained in 64*64 matrices. In clinical practices about 25 mCi of ^{99m}Tc is injected into patient.

Results

A. MONTE CARLO SIMULATED PHANTOMS

Simulations were carried out with the number of simulation 10^5 and the activity 2mci to make a high noise bar phantom .For all of images in this study we experimentally chose D4 Wavelets at level 2 .Also for butterworth filteration we chose butterworth filter with order 7 and cutoff frequency of 0.45 recommended in some studies[6]. Figures 4-5, shows the results of Power Spectrum calculations for cold and hot phantom which show better denoising with platelet-based MPLE in compared with wavelet and butterworth filteration. As we see high frequency noise was removed well according to the original power spectrum whereas it remained in resulting image for butterworth filteration . Table 1 shows obtained values for noise level measurements for hot and cold phantoms.

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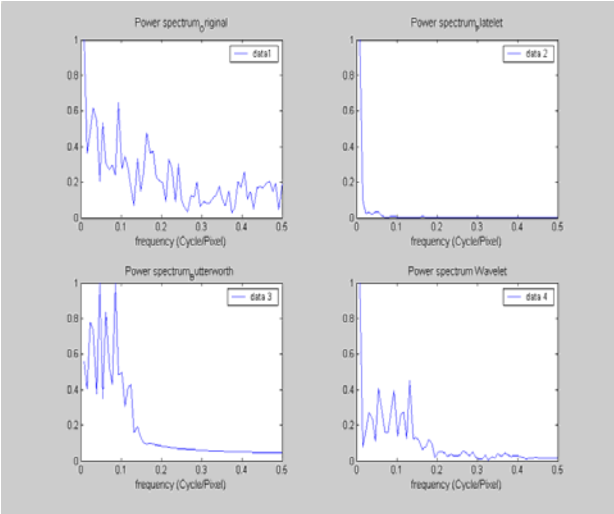


Figure 4: Calculated Power Spectrum for reconstructed phantom (original) and denoised with Platelet-based MPLE , Wavelet and Butterworth filteration of cold phantom.

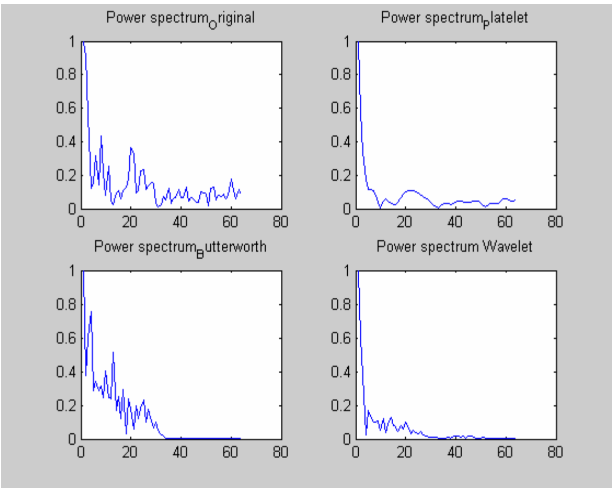


Figure 5: Calculated Power Spectrum for reconstructed phantom (original) and denoised with Platelet-based MPLE , Wavelet and Butterworth filteration of hot phantom.

Table 1: Noise level measurements for hot and cold simulated phantoms

	Original	Butterworth	Platelets	D4 wavelets
Cold	0.0733	0.0647	4.327e-004	0.0027
Hot	0.0698	0.0528	3.602e-004	0.0024

B. NEMA Phantom

For 32 images of NEMA phantom we calculated the Mean and Standard Deviation of noise level .we chose D4 Wavelets at level 2. for butterworth filteration we chose butterworth filter with order 7 and cutoff 0.45 recommended in some studies[6].

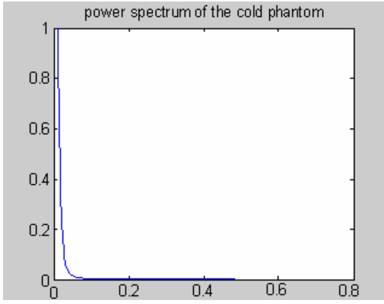


Figure 6: Calculated Power spectrum for simulated cold and hot phantom(original object).

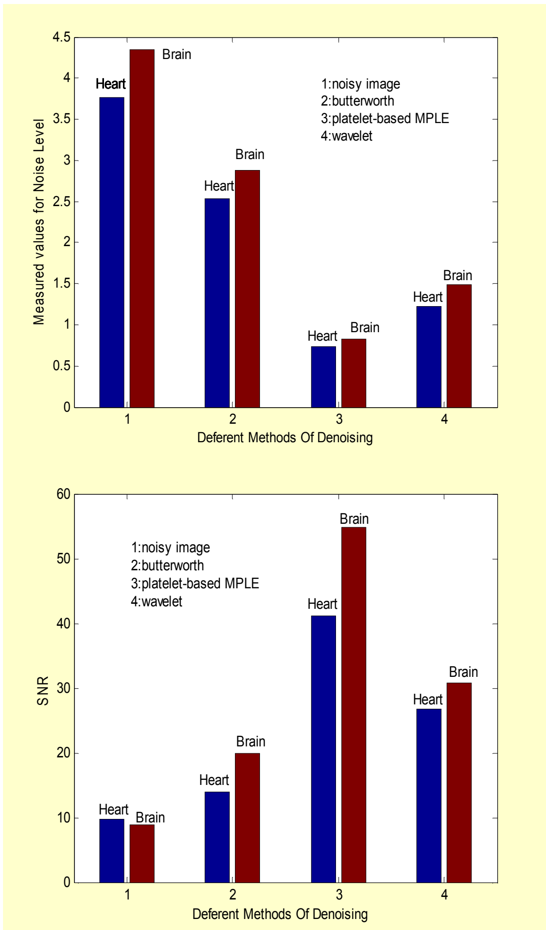
Table 2: Results for NEMA images

	Original	Butterworth	Platelets	D4 wavelets
Mean of noise level	2.1732	1.3566	0.1399	0.6770
SD	±0.1711	±0.0969	±0.0256	±0.0556

C. Patient Study

For 32 cardiac SPECT images and 32 brain SPECT images following results for measured noise levels were obtained.

Table 3: Results of Measured noise levels and SNR values for cardiac and brain SPECT images .



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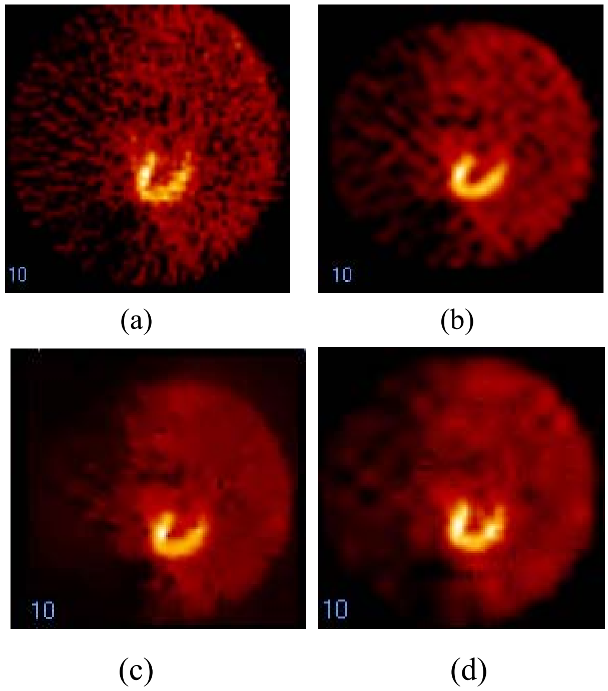

Figure 7: Cardiac SPECT image denoised with 3 techniques: (a) initial ; (b)butterworth ; (c) platelet-based MPLE ; (d) wavelet

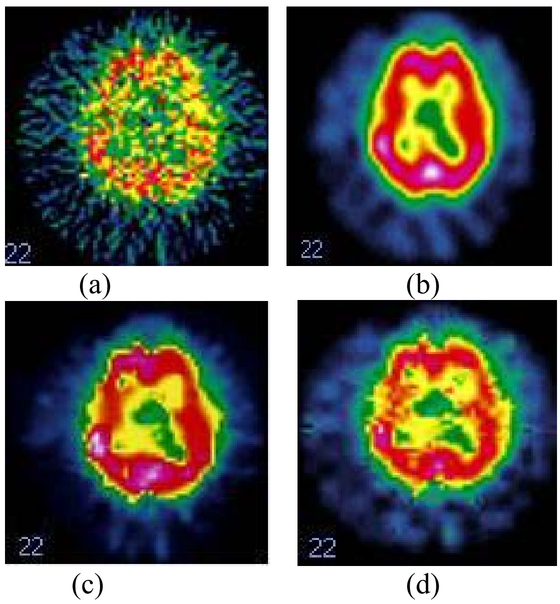

Figure 8: Brain SPECT image denoised with 3 techniques: (a) initial ; (b)butterworth ; (c) platelet-based MPLE ; (d) wavelet

Table 4: Clinical evaluation of cardiac images by nuclear medicine specialists

Original	Butterworth	Platelets-based MPLE	D4 wavelet
Very poor	good	Excellent	Acceptable

Table 5: Clinical evaluation of brain images by nuclear medicine specialists

Original	Butterworth	Platelets-based MPLE	D4 wavelet
Very poor	Excellent	good	Acceptable

Conclusion

Our results showed that Platelet-based MPLE as a multiscale decomposition method is useful for denoising SPECT images considering improvements in representing edges and smooth regions , better homogenous image, better representing radioactive uptake in target organ and reduction of interfering activity from background radiation (statistical fluctuation) in compare with that of other conventional denoising methods.

Wavelets and Butterworth filteration techniques do not perform well on this class of images with inherent smoothness and Poisson noise due to the quantum nature of the photon detection process .

Appendix

A. Reviewing Platelet Analysis

Platelets are localized atoms at various locations, scales and orientations that can produce accurate, piecewise image approximations to images consisting of smooth regions separated by smooth boundary[3].

In Haar analysis the image is approximated on each cell of the partition by a constant. in platelet analysis it can be approximated with a planar surface by defining a platelet

f(x,y) to be a function of the form

$$f_s(x,y) = (A_S x + B_S y + C_S) I_S(x,y) \qquad (2)$$

where $A_S, B_S, C_S \in$ IR, S is a dyadic square associated with a terminal node of an RP(Recursive Partition), and I_S denotes the indicator function on S. Each platelet requires three coefficients, compared with the one coefficient for piecewise constant approximation.

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B. Image Denoising

By adding a penalizing function to the log-likelihood function in maximum likelihood criteria, Maximum Penalized Likelihood criteria is produced. The penalizing function measures the smoothness of the intensity image and assigns weaker penalties to smoother intensities and stronger penalties to more irregular intensities.

Goal here is to maximize the penalized likelihood function

$$L_{\gamma}(\mu) \equiv \log p(x/\mu) - \gamma\{\neq \theta\} \qquad (3)$$

where p(x|μ) is the likelihood factorization ; γ is a smoothing parameter and {#θ} is the number of parameters in the vector θ. So, the MPLE is as following equation:

$$(\hat{P}, \hat{\theta}) \equiv \arg\max_{p,\theta} L_{\gamma}(\mu(P,\theta)) \qquad (4)$$

Which yields a maximum penalized likelihood estimator (MPLE). Larger values of γ produce smoother, less complex estimators; smaller values of γ produce more complicated estimators.

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