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A Quantitative Comparison between Butterworth and Metz Filter Applied In Quantitative Spect Myocardial Perfusion Imaging Study --- A Retrospective Evaluation

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Abstract
Aims and objective: *Quantitative comparison of the efficacy between Butterworth and Metz filters applied in quantitative Myocardial SPECT perfusion study.*
Materials and Methods: *An algorithm has been established, based on 17 segment model with the help of which the site of coronary artery obstruction can be localised . .Each of the raw myocardial SPECT images of the 30 patients under study were filtered before reconstruction by Butterworth and Metz filter respectively. Thereafter, two sets of algorithm from 17 segment model were generated from each of the two sets of filtered images from every patient. ,since coronary angiography is still being considered as the gold standard to localize anatomical position of coronary arterial obstruction, we have compared the calculated results obtained, following application of algorithm to both Butterworth filtered image and Metz filtered image with the angiographic findings.*
Results: *Although both Metz and Butterworth filtered image fit closely with angiographic findings, it is the Butterworth filter that matches the best , with angiography.*
Conclusion: *The superior efficacy of Butterworth filter as compared to Metz filter has been proved by quantitative method through this study,in contrast to most of the previous studies which were based solely on subjective comparison*

Keywords: *Modulation Transfer function (MTF), Point spread function (PSF), Full width half maximum(FWHM), Nyquist Frequency(NF), signal-Noise Ratio (SNR), Fourier transformation*

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I. Introduction:

Nuclear medicine images are used for both qualitative and quantitative assessment. But the qualitative assessment of the images is not as reliable as the quantitative interpretation.This is due to the reason that mere visual interpretation of an image has limitation of inter and intra subjective variation as well as long time consumption. on the other hand the quantitative study provides an objective interpretation that is more reproducible than visual analysis.

For an easier and more objective assessment of myocardial perfusion, cardiac polar plot has been developed in cardiac SPECT which consists of single image display of the information contained in the short axis slices of the left ventricle. The left ventricle is divided into 17 segments popularly known as 17 segment model where each of these 17 segments is analysed with five point perfusion system[1].

Although myocardial perfusion defect can be detected through cardiac SPECT, it has not yet been possible to localize the anatomical location of obstruction in the coronary artery by this mode of imaging. The reason for the failure of this image system may be due to the fact that SPECT imaging is a functional imaging and there is lot of individual variation of coronary arterial anatomy., That is why coronary angiography is still being considered as the gold standard for the purpose of localization of arterial obstruction.

Based on the 17 segment model in myocardial perfusion study by SPECT.,Pal, Sushanta, et.al.and Basu, Sandip have invented an algorithm[2] to determine the anatomical location of coronary arterial obstruction,through SPECT myocardial perfusion study. Both the quality of the image and size precision of the defect are the absolute requirements for accurate quantitative analysis from an image.

Additional image processing is needed for this purpose that ensures improvement of quality of the image and size precision of the defect through better resolution, reduction of artifact, contrast enhancement as well as noise reduction[3].

All forms of imaging are badly affected by statistical variation in the acquired image counts, commonly referred to as noise. Generally, the more the noise poorer is the imaging quality. Compared to planer images SPECT images have fewer counts therefore always have more noise. As the main area of concern for an interpreter of SPECT studies is image enhancement by reducing noise prior to image reconstruction , more accurate filter is always being looked for, so that the purpose of better elimination of unwanted data and enhancement of true counts can be achieved.

In SPECT, image filtering requires the conversion of data from spatial to the frequency domain, using the Fourier transform. The Fourier transform of the convolution is the product of the two Fourier transforms[11] – This is the Convolution Theorem. Deconvolution is an algorithm based process used to enhance signals from the recorded data,used to restore the original signal when a recorded data can be modelled as a pure signal distorted by a filter through the process known as convolution. The convolution process in the spatial domain is equivalent to a multiplication process in the spatial frequency domain[7],

In medical imaging the amplitude versus Distance plot can be considered as count rate profile through an image of a point source of radioactivity, termed as a point spread function(PSF) and its Fourier spectrum is called .Modulation Transfer function (MTF). The image restoration process is actually an enhancement or suppression of certain spatial frequencies in the Fourier spectrum by using filters. The filtered spectrum can then by the process of inverse Fourier transform , generate a filtered image with smoothed features.Filters are applied to projection images before reconstruction to reduce noise early in the processing chain. In a SPECT study by setting the correct Nyquist frequency (NF) we can reconstruct the entire spectrum of spatial data. A lot of different wave patters are found within NF. In general frequencies can be broken down into three categories.

Low - background and large objects
Middle - variation of smaller objects
High frequencies - small to very small objects plus noise.

It is very difficult to differentiate between very small objects and noise because it goes beyond the resolution of the imaging system (FWHM).

It has been found that 99mTc images contain significant features in the intermediate frequency ranges. Both the amplitude of data of the concerned organ image as well as amplitude of noise vary differently with frequency and total count of the acquired image. The amplitude of noise remains constant in all frequencies. In contrast amplitude of data of broad features of the organ of interest reduces with the high frequency and vice versa. Therefore at low frequency, object data dominates the noise .In contrast at high frequency data disappears in the noise. So as the image count (%uptake) increases, high frequency components due to random noise decrease and the object frequencies can be separated from the noise at high frequencies[8]

Filtering before back projection involves Ramp , a high pass filter that sharpens the edges of an image and enhances object edge information. Since this filter amplifies the high frequencies, resulting in more noise and consequent degradation of signal-to-noise ratio(SNR i.e the relative strength of the signal component that is actually being imaged, compared to noise[5].), another smoothing filter is needed to counteract this effect .The two important smoothing filter parameters are cut off frequency and order.. cut-off frequency determines where the filter rolls off to zero gain. A low cut off frequency increases smoothing but degrades contrast. A high cut off frequency improves the spatial resolution at the cost of statistical noise. Another parameter that is a determining factor is the order or power. The power factor controls the steepness of roll off.[6] .Higher the order sharper is the fall of slope[1]..

Aims and Objective :
The aim of our study is to find out the most appropriate filter, objectively for the purpose of quantitative SPECT myocardial perfusion study between Butterworth and Metz filter.

II. Material And Methods :

This retrospective study .included thirty patients, who had angiographic evidence of more than 50% stenosis and SRS score of more than or equal to 13 .Rest gated image was taken 60 minutes after injection of 99mTC sestamibi. Prone imaging in addition to supine imaging, was taken when necessary, to eliminate subdiaphragmatic and breast attenuation.

Dual head gamma camera(Siemens)equipped with high resolution collimator has been used A protocol consisting of 64x64 matrix , 32 projection per head ,20 second projection and 8 frames per cycle was applied with appropriate energy photo peak..

Coronary angiography was performed within two weeks after gated SPECT in 90% case and in 10% cases coronary angiography had been performed before gated SPECT .No patients had experienced any cardiac symptoms or angiographic changes within this interval of time between gated SPECT and angiogram .The raw SPECT cine images were viewed for potential sources of artefact before interpreting the reconstructed myocardial perfusion study.

Each raw image was subjected to filtering separately with Butterworth filter and Metz filter. Individual filter parameters were applied to the raw image using the serial filter software. We had chosen two separate sets of optimal filter functions, for both Butterworth and Metz filter. Butterworth filter ,with an order of 5 and cut off frequency of 0.45 and Metz filter with FWHM value of 19 and Order of 5 have been selected for the study . Metz filter has two parameters to be adjusted, MTF and order. In our system FWHM is used instead of MTF and the system itself calculates the MTF. .

We selected the filter parameters by applying them to the respective filters by trial and error method and simultaneously noting the gain in the gain frequency curve of the applied filter and resultant subjective variation of the quality of the image. In our software minimum FWHM was set at 6 mm and order as 1. In case of Metz filter . We put the FWHM and order in such a way that the gain in the gain versus frequency curve is accentuated to 1.5 within the middle frequency range .in case of Butterworth filter it was maintained at 1 and the filter asymptotically approaches to zero in order to exclude the higher frequencies..

The Corridor4DM (4DM) application (formerly known as 4DM-SPECT) a quantitative software has been used for the analysis and review of reconstructed myocardial image. The SRS score by the perfusion scoring system in the 17 segment model was then calculated . Only those patients who had had scored SRS(Summed Rest Score) value equal to or more than 13 by the perfusion scoring system in 17 segment model were included in the study.

The location of coronary stenoses was categorized as proximal, middle, or distal for left anterior descending artery and as proximal or distal for left circumflex and right coronary arteries. If there are more than one stenosis in any coronary artery, only the most severe one would be taken into consideration ignoring the eventual; downstream lesions .

The 17-segment model provides the best agreement with the available anatomic data and has the best fit with the methods commonly used in both echocardiography and SPECT nuclear cardiology.

An additional analysis, including adjacent segments, demonstrated that the coronary artery territory can be more accurately localized[4]

For instance the inferior and inferolateral segments can belong to either RCA or the LCX artery. However if the anterolateral segments are also involved in a given patient the distribution appears to be more specific for the LCX, whereas if the inferoseptal segments are involved , the distribution appears to be more specific for the RCA. Similarly inferoseptal segments can belong to either the LAD or RCA but if the adjacent inferior segments are also involved the territory appears to be more specific for the LAD . likewise the inferoapical segment can belong to either the RCA or LAD but if the other apical segments are involved the territory appears to be more specific for the LAD. if the inferior segments at the mid and basal levels are involved the distribution appears to be more specific for RCA. the greatest variability in myocardial blood supply occurs at the apical cap, segment 17 which can be supplied by any of the three arteries. segments 1,2,7,8,13,14, and 17 are assigned to the LAD.segment 3,4,9.10 and 15 are assigned to theRCA when it is dominant. segments 5,6,11, 12 and 16 generally are assigned to the left circumflex artery. each segment was scored automatically by the use of software based on a5 point scoring system(0, normal uptake;1, mild decreased uptake ;2, moderate decreased uptake ; 3, severely decreased uptake 4, absence of uptake).

The final score (ranging from 0 to 17) was obtained by summation of the perfusion score of the individual segments belonging to specific coronary artery as determined by the aforesaid analysis.

summed scores of 7,5, and 3 in the LAD domain were assigned to proximal, middle and distal part of stenosis respectively. The score of 5 and 3 in both left circumflex and right coronary artery domains were assigned to proximal and distal stenosis respectively in both left circumflex and right coronary artery. A score of 12 was assigned to left main stenosis .

III. Results:

Through data analysis three sets of 2*2 contingency tables were generated

1. comparing angiographic data specifying anatomical location of arterial obstruction along the row versus specified anatomical locations of arterial obstruction as per the algorithm generated from the Metz filtered reconstructed image , along the column.
2. comparing angiographic data specifying anatomical location of arterial obstruction along the row versus specified anatomical location of arterial obstruction as per the algorithm generated from the Butterworth filtered reconstructed image , along the column.

3. specified anatomical locations of arterial obstruction as per the algorithm generated from the Metz filtered reconstructed image along the column and specified anatomical location of arterial obstruction as per the algorithm generated from the Butterworth filtered reconstructed image .

The Kappa statistic (or value) is a metric that compares an Observed Accuracy with an Expected Accuracy (random chance).:unweighted Kappa value of the comparative data between butterworth filtered data versus angiographic data is K=0.803which is suggestive of substantially strong agreement. Unweighted Kappa value of the comparative data between Metz filtered data versus angiographic data is K=0.576 .This suggests moderate agreement:Unweighted Kappa value of the comparative data between butterworth filtered data and Metz filtered data. Is K=0.623,suggestive of strong agreement.

Kappa result be interpreted as follows: values ≤ 0 as indicating no agreement and 0.01–0.20 as none to slight, 0.21–0.40 as fair, 0.41– 0.60 as moderate, 0.61–0.80 as substantial, and 0.81–1.00 as almost perfect agreement.

IV. Discussion :

The quality of an image can be described as the SNR , The SNR is much higher at lower spatial frequencies (broad features that are constant over many pixels) and decreases at higher spatial frequencies (features that change over few pixels such as edges).. SNR varies with count statistics directly . Although resolution and contrast are close concepts but they are different in nature.

Resolution is defined as the ability to separate close distance structure and it reflects ability to show details and represented by highest frequency component of the data in the frequency domain. Contrast can be calculated by subtracting the minimum count in the defect from the maximum count in normal myocardium and the result is divided by the maximum count in normal myocardium.

The concept of restoration filter is to attempt to recover the resolution lost in the detection process on a low count by restoring counts in an image. Hence, images with low count density may benefit with this type of application exceeding unity gain over a desired middle frequency data. But because of domination of noise at higher frequency this filters at some point must roll off to zero gain .Metz filter is a combination of deconvolution and smoothing filter and it has got a parameter that determines the extent to which the filter follows the inverse modulation transfer factor(inverse-MTF) before rolling off to zero gain[6]

MTF is a the normalised Fourier transform of the point spread function(PSF)[13]

Metz filter is a deconvolution procedure assuming knowledge of the point spread function of the imaging system[14].image restoration by by deconvolution can significantly improve spatial resolution of an image improving visibility of small structures. The quality of a degraded image can be significantly restored by deconvolution with PSF.[15]. So it can be said that spatial resolution of the restored image by Metz filter can be significantly improved accuracy of the size of the lesion image. But a significant improvement of image quality can also be obtained by high pass filtering using a butterworth filter with a cutoff frequency matching that of the PSF.

Metz filter works similar to Ramp in that it amplifies frequencies but an inverse MTF causing the amplification of low and mid frequencies instead of higher frequencies. It may be considered as the product of inverse filter(which shows resolution recovery) and low pass filter (which shows noise suppression).

Metz filter is exclusively a function of MTF and has basically nothing that is related to signal noise ratio . It has been observed that the frequency curve(i.e MTF of the gama camera) gets interrupted in the middle by low count noise curve but for higher count the noise level breaks (away from the MTF curve into higher frequency range. so for higher count images MTF curve is not significantly distracted. It is impossible to find the difference between small objects and noise, Once the noise curve breaks from the MTF curve. Therefore for an ideal image Metz filter becomes neutral. Although it is implied that MTF is a measure of signal noise ratio, that is not necessarily true when statistics is poor or scatter is high.

Resolution reflects ability to show details and is represented by highest frequency component of the data[12]. But Metz filter enhances some middle frequency data .although increases some component of the image but due to inverse MTF effect higher frequencies are missed ,so they cant be truly regarded as resolution enhancement filter.

In contrast for Butterworth Filter, cut-off frequency has been defined as the point where the gain is down to 0.707..Butterworth filter maintains a unit gain at lower frequencies and Metz filter exceeds a unit gain in lower frequencies. But in case of Butterworth filter with our prescribed filter parameters, gain frequency curve asymptotically approaches zero thus recruiting important minute details of the of the organ that lies in the higher frequency zones. So due to this unique curve pattern of Butterworth filter both SNR as well as finer details of the organ image are in better compared to Metz filter .

Another factor necessary for quantitative study is contrast enhancement. Defect becomes prominent when defect contrast is enhanced[9].

..data shows that calculation of maximum contrast between normal and defected myocardium could be obtained using Metz,. Result of Metz filter as compared to butterworth should better match with angiography because the filter that produce maximum contrast is assumed to more consistent with angiography[10.]... Between butterworth and Metz filter , the later increases contrast maximum at some specific frequencies.

On the other hand, a profile through the lesion centre from a projection image in which lesion is at a distance from the detector approximately equal to the radius of rotation shows that Butterworth filter with the said parameters is very close in size to the detected lesion compared to Metz filter[6] . The principle of Metz filter is based on generalised deconvolution scheme developed

Originally for conventional radionuclide image processing. It shows little ringing because of Gibbs phenomenon. Gibbs phenomenon reflects the difficulty, inherent in approximating a discontinuous function by a finite series of continuous sine and cosine waves. The frequency domain perspective is, that little ringing is caused due to sharp cut off in the reactangular pass band in the frequency domain. On the contrary in case of Butterworth filter, in the , filter roll-off is the slowest, and smoother cutoff in the rectangular pass band results in the fewest time domain artefact but leaks most in the stop band. So Size accuracy seems better in Butterworth compared Metz filter

V. Conclusion :

it can be concluded that, although Butterworth and Metz filtered images are comparable for quantitative study of SPECT myocardial perfusion images ,we would choose Butterworth for this purpose. This is due to the fact that although the mean contrast,SNR and defect size are not better compared to that of Metz , the mean total grade of these three factors is higher in Butterworth compared to Metz.This, plays the major determining factor in selecting the fiter that would help the quantitative model to approximate angiography..

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		Anatomical location of Coronary obstruction data derived from Angiography							
Obstruction location obtained through calculation derived from Metz filtered data using algorithm		$LAD_{0/P}$	$LAD_{M/D}$	LCX_P	LCX_D	RCA_P	RCA_D	LCX_0	RCA_0
	LAD0/P	14	5	0	0	0	0	0	0
	$LAD_{M/D}$	9	5	0	0	0	0	0	0
	LCX_P	0	0	13	2	0	0	1	0
	LCX_D	0	0	0	0	0	0	2	0
	RCA_P	0	0	0	0	9	3	0	0
	RCA_D	0	0	0	0	2	2	0	0
	LCX_0	0	0	1	0	0	0	11	0
	RCA_0	0	0	0	0	0	3	0	6

Table 1. Anatomical location of coronary obstruction data derived from angiography versus same obtained from calculation usingalgorithm from Metz filtered image.

n= 94

Unweighted k = 0.576

95% C.I = 0.465 to 0.688

		Anatomical location of Coronary obstruction data derived from Angiography							
Obstruction location obtained through calculation derived from Butterworth filtered data using algorithm		LAD_0	LAD_P	LADM/D	LCX_P	RCA_P	RCA_D	LCX_0	RCA_0
	LAD_0	6	0	0	0	0	0	0	0
	LAD_P	0	16	2	0	0	0	0	0
	$LAD_{M/D}$	1	1	8	0	0	0	0	0
	LCX_P	0	0	0	13	0	0	1	0
	RCA_P	0	0	0	0	12	1	0	1
	RCA_D	0	0	0	0	2	2	0	5
	LCX_0	0	0	0	1	0	0	14	0
	RCA_0	0	0	0	0	0	1	0	7

Table 2. Anatomical location of coronary obstruction data derived from angiography versus same obtained from calculation using algorithm from Butterworth filtered image.

Unweighted k = 0.80

95% C.I 0.716 to 0.890

	Obstruction location obtained through calculation derived from Butterworth filtered data using algorithm							
Obstruction location obtained through calculation derived from Metz filtered data using algorithm		LAD_0	LAD_P	LAD_M	LAD_D	LCX_P	RCA_0	RCA_P	RCA_D
	LAD_0	4	0	0	0	0	0	0	0
	LAD_P	0	11	2	1	0	0	0	0
	LAD_M	0	4	2	0	0	0	0	0
	LAD_D	4	3	0	3	0	0	0	0
	LCX_P	0	0	0	0	16	0	0	0
	RCA_P	0	0	0	0	0	9	2	0
	RCA_D	0	0	0	0	0	2	5	7
	RCA_0	0	0	0	0	0	1	1	6

Table 3. Anatomical location of coronary obstruction data derived from calculation using algorithm from Butterworth filtered image versus same obtained from calculation using algorithm from Metz filtered image.

Unweighted k = 0.623

95% C.I 0.509 to 0.736

	Anatomical location of Coronary obstruction data derived from Angiography								
Obstruction location obtained through calculation derived from Metz filtered data using algorithm		$LAD_{0/P}$	$LAD_{M/D}$	LCX_P	LCX_D	RCA_P	RCA_D	LCX_0	RCA_0
	$LAD_{0/P}$	14	5	0	0	0	0	0	0
	$LAD_{M/D}$	9	5	0	0	0	0	0	0
	LCX_P	0	0	13	2	0	0	1	0
	LCX_D	0	0	0	0	0	0	2	0
	RCA_P	0	0	0	0	9	3	0	0
	RCA_D	0	0	0	0	2	2	0	0
	LCX_0	0	0	1	0	0	0	11	0
	RCA_0	0	0	0	0	0	3	0	6

Table 1. Anatomical location of coronary obstruction data derived from angiography versus same obtained from calculation usingalgorithm from Metz filtered image.

n= 94

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Obstruction location obtained through calculation derived from Butterworth filtered data using algorithm		LAD_0	LAD_P	$LAD_{M/D}$	LCX_P	RCA_P	RCA_D	LCX_0	RCA_0
	LAD_0	6	0	0	0	0	0	0	0
	LAD_P	0	16	2	0	0	0	0	0
	$LAD_{M/D}$	1	1	8	0	0	0	0	0
	LCX_P	0	0	0	13	0	0	1	0
	RCA_P	0	0	0	0	12	1	0	1
	RCA_D	0	0	0	0	2	2	0	5
	LCX_0	0	0	0	1	0	0	14	0
	RCA_0	0	0	0	0	0	1	0	7

Table 2. Anatomical location of coronary obstruction data derived from angiography versus same obtained from calculation using algorithm from Butterworth filtered image.

Unweighted k = 0.80

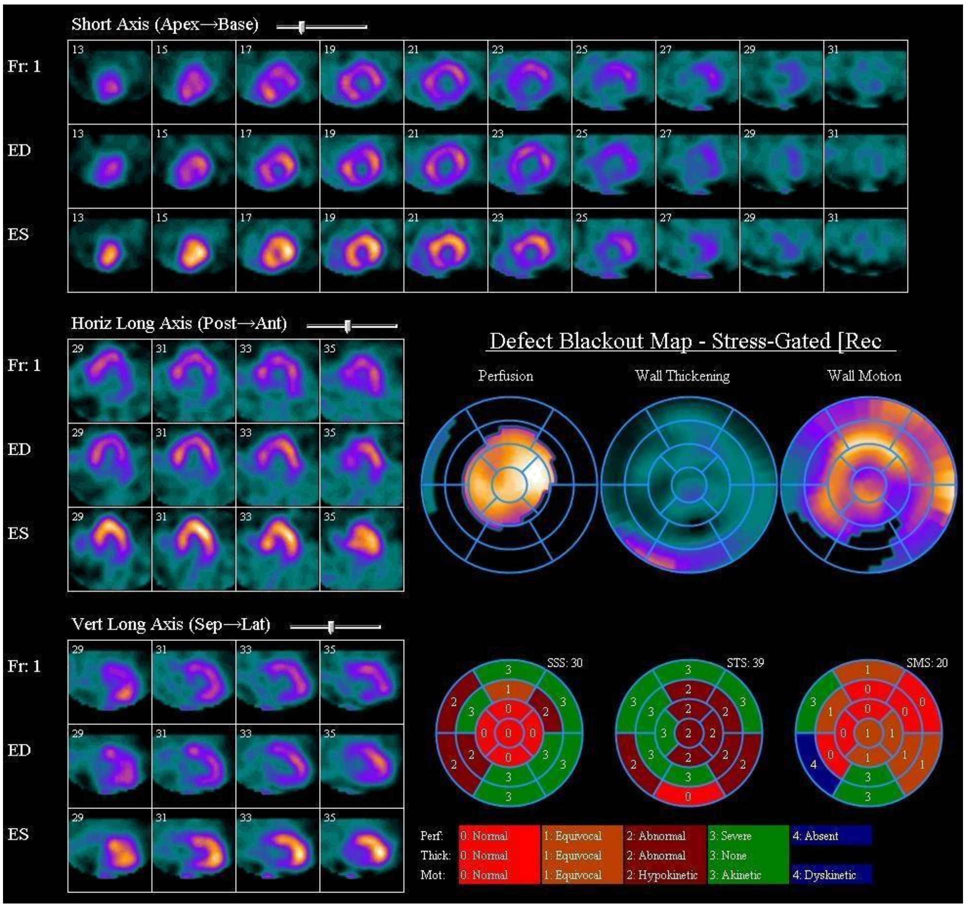
95% C.I 0.716 to 0.890

		Obstruction location obtained through calculation derived from Butterworth filtered data using algorithm							
Obstruction location obtained through calculation derived from Metz filtered data using algorithm		LAD_0	LAD_P	LAD_M	LAD_D	LCX_P	RCA_0	RCA_P	RCA_D
	LAD_0	4	0	0	0	0	0	0	0
	LAD_P	0	11	2	1	0	0	0	0
	LAD_M	0	4	2	0	0	0	0	0
	LAD_D	4	3	0	3	0	0	0	0
	LCX_P	0	0	0	0	16	0	0	0
	RCA_P	0	0	0	0	0	9	2	0
	RCA_D	0	0	0	0	0	2	5	7
	RCA_0	0	0	0	0	0	1	1	6

Table 3. Anatomical location of coronary obstruction data derived from calculation using algorithm from Butterworth filtered image versus same obtained from calculation using algorithm from Metz filtered image.

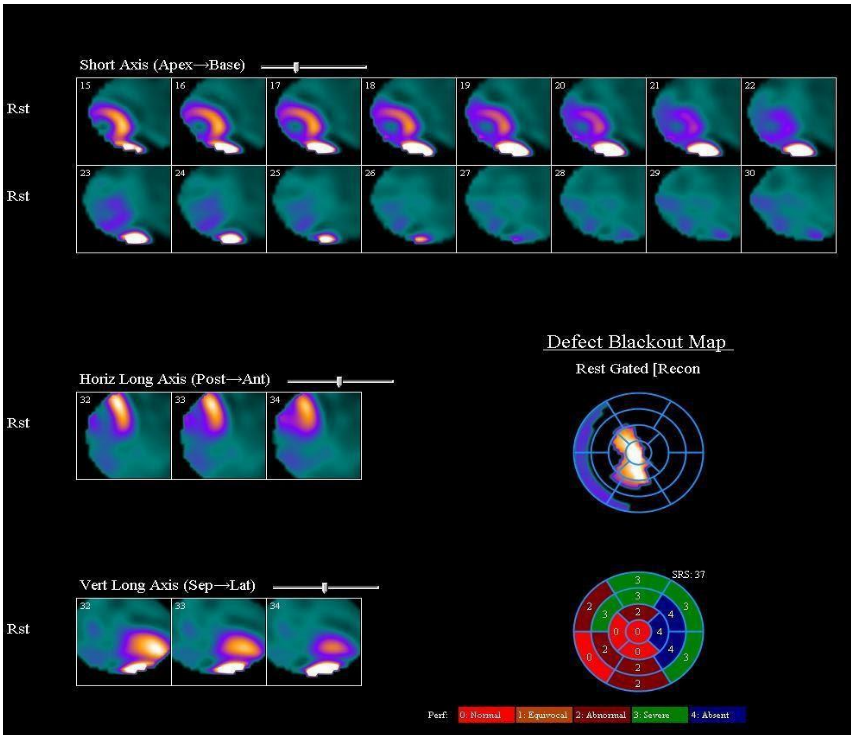
Unweighted k = 0.623

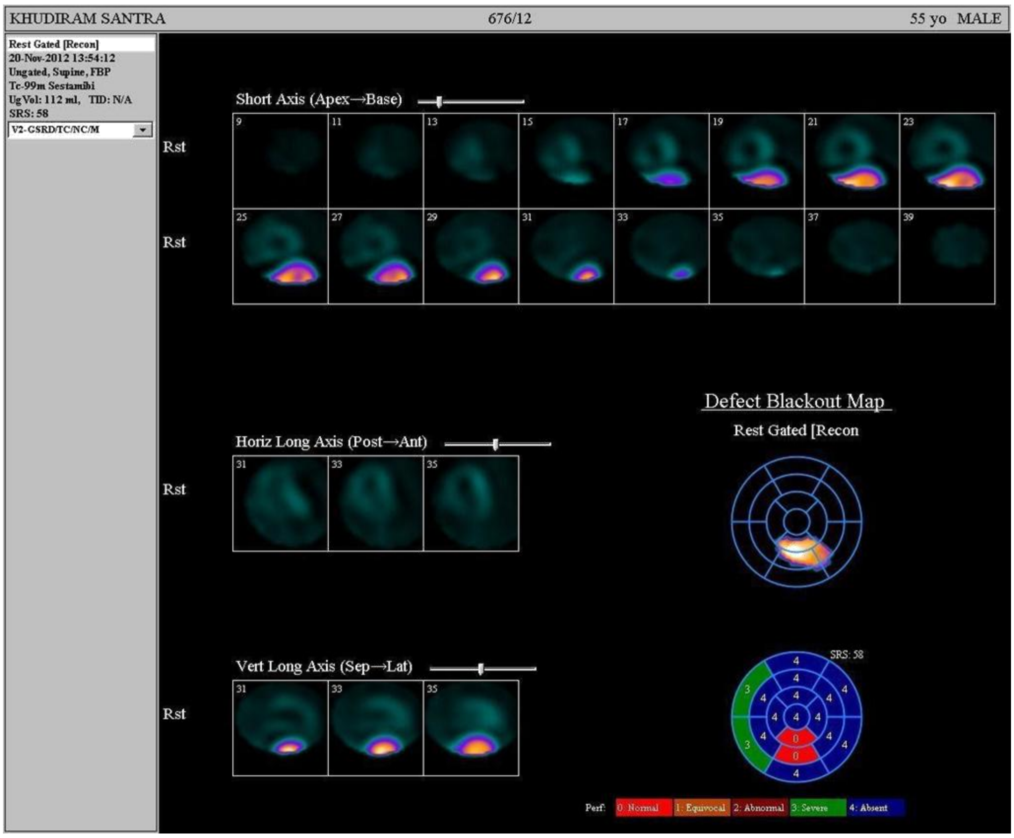
95% C.I 0.509 to 0.736



A QUANTITATIVE COMPARISON BETWEEN BUTTERWORTH AND METZ FILTER APPLIED IN QUANTITATIVE SPECT MYOCARDIAL PERFUSION IMAGING STUDY---A RETROSPECTIVE EVALUATION

MEDICINE





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