EVALUATION OF VARYING ACQUISITION PARAMETERS ON THE IMAGE CONTRAST IN SPECT STUDIES

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ABSTRACT

Many physical factors degrade SPECT images both qualitatively and quantitatively. SPECT has highlighted the need to improve gamma camera performance. The study will examine different acquisition parameters for the gamma camera such as matrix size and motion type. Also it will cover the processing parameters like filtered back projection and iterative reconstruction techniques.

Image contrast refers to the differences in density (or intensity) in parts of the image corresponding to different levels of radioactive uptake in the patient. In nuclear medicine, it is the lesion-to-background uptake or concentration ratio. Various physical factors limit the quality of the SPECT images. Among these are matrix size and motion type. A general definition of contrast is that it is the ratio of signal change of an object of interest, such as a lesion, relative to the signal level in surrounding parts of the image.

Aim of the study: The aim of the present work is to evaluate different physical factors (matrix size & motion type) affecting the image contrast in Single Photon Emission Computerized Tomography (SPECT) in order to optimize the patient's examination on the gamma camera in Nuclear medicine facility.

Key words: *SPECT, physical factors, contrast, FBP, Iterations, Gamma Camera*

1. INTRODUCTION

SPECT (Single Photon Emission Computed Tomography) is a technique for tracing radioactive tracers in the body and provide three dimensional maps of in vivo radiopharmaceutical distributions [(1)]. Single Photon Emission Computed Tomography technique used to image cross section planes of the target organ, its main benefit is:
Increase the image contrast due to reduction in background activity super imposed on object activity, gives depth information about the deep laying lesions, and to increase the accuracy of quantification.

If one rotates the camera around the patient, the camera will acquire views of the tracer distribution at a variety of angles. After all these angles have been observed, it is possible to reconstruct a three dimensional view of the radiotracer distribution within the body. The fundamental goal of all tomographic imaging systems is to portray the distribution of the radioactivity in the patient more accurately [(2)].

SPECT imaging is superior to planar imaging with regard to disease localization by improving the object contrast [(3)]. The degradation of planar images in nuclear medicine is usually easy to recognize in clinical images obtained from a poorly functioning scintillation camera. This is not usually the case in SPECT where it is quite possible to produce poor images or images containing artifacts without the situation being properly recognized. There are whole ranges of acquisition and processing parameters that can affect the quality of the final image [(4)].

Physical properties important for the assessment of the potential of emission computed tomography implemented by collimated detector systems include sensitivity, statistical and angular sampling requirements, attenuation compensation, resolution, uniformity, and multi-section design constraints. SPECT has highlighted the need to improve gamma camera performance. Flood field non-uniformity is translated into tomographic images as major artifacts because it distorts the data obtained at each projection. Also, poor energy resolution translates directly into degraded spatial resolution through reduced ability to reject scattered photons on the basis of pulse height analysis [(5)].

Image contrast refers to the differences in density (or intensity) in parts of the image corresponding to different levels of radioactive uptake in the patient. In nuclear medicine, it is the lesion-to-background uptake or concentration ratio.

A general definition of contrast is that it is the ratio of signal change of an object of interest, such as a lesion, relative to the signal level in surrounding parts of the image. For multi-head systems, the tomographic contrast is an important indicator of how well a system is performing with respect to detection of small lesions. Here it is defined as follows; Place a sphere of some known size within a volume containing a uniform concentration of activity. After reconstruction, estimate the value (***C bgd***) for background count of pixels for the reconstructed image in the neighborhood of the sphere, but outside the region corresponding to the sphere. Estimate also the value of pixels within the region corresponding to the sphere (***C sph.)*** for sphere count**.** Contrast for this size lesion may then be calculated as:

$$\textbf{Contrast} = \frac{(C\ sph. - C\ bgd)}{(C\ sph. + C\ bgd)}$$

Many other possible definitions exist and have been employed. However, the fundamental concept is to estimate the ability of the system to detect a known change in activity concentration, for a given size of a spherical object. In particular, contrast is very dependent on the size of the lesion used to estimate it. Tomographic contrast is important in that it determines the detectability of small lesions. It is affected by many different physical factors that limit the quality of the SPECT images.

2. MATERIALS AND METHODS

2.1. Materials

The gamma camera used in the present study is Philips (Axis) dual head, the number of photomultiplier tubes is 54, the detector UFOV dimensions are (55 X 40) cm. The camera computer system is a LUNIX based system with different software attained from different institutions such as EMORY, SEDER –SINAI and European software programs for processing and displaying, different cases such as static, dynamic, total body and SPECT. There are also programs for manipulation of the images and for quality control calculations which are used in this study such as image profile, region of interest and pixel count calculation. There are also quality control programs that are used for calculation of the center of rotation of the camera and pixel calibration for each mounted collimator to the camera heads. The reconstruction algorithms used for SPECT reconstruction are both filtered back-projection (FBP) and iterative reconstruction algorithms.

All images were acquired using a SPECT phantom (Jasezczak phantom). This phantom is designed from clear acrylic source tank which can be filled with a Tc-99m and water solution similar to that used for routine flood uniformity testing. This tank contains a set of three inserts includes two for resolution (one with “cold” lesions in a “hot” field and one with “hot” lesions in a “cold” field) and one for linearity/uniformity measurements .

It also contains eight pairs of holes drilled through a solid acrylic block. The hole diameters are4.7, 5.9, 7.3, 9.2, 11.4, 14.3, 17.9, 22.3 and 38.1mm. The diameter of each pair increases nominally by 25% over that of the preceding pair. The solid block creates a “cold” field in which the solution-filled holes appear as “hot” lesions. In this study we will use this phantom to measures the contrast of the image, using those hot spheres in the cold background.

The radioisotope used for imaging was Technetium (Tc^{99m}) .The Elutec Technetium (Tc^{99m}) generator produces a sterile solution of Tc^{99m} as sodium pertechetate .This solution is eluted using a0.9% sterile and endotoxin free solution of sodium chloride from an alumina chromatography column to which the Mo^{99}($T^{1/2}$ =66.02h) parent of (Tc^{99m}=6.02h) is adsorbed.

2.2. Methods

The SPECT phantom filled with water mixed with (25 mCi) of $Tc^{99m,}$ was positioned on a special holder attached to the imaging table. The cylinder axis of the phantom was parallel to the axis of rotation of gamma camera detector, within the rotational Useful field of view.

After acquiring the raw image for each test at different value for each parameter, those images were reconstructed image reconstructed by 2 reconstruction methods (Filtered Back Projection & Iterative).Using Butterworth filter with cutoff value = 0.5 and order value = 4.0, with the reconstruction parameters which is the same reconstruction parameters applied on the clinical cases.

Then the contrast of image was calculated by the equation (1) after measuring the count in one of the sphere and the background count for a sphere with the same size in the background region, and away from the edges to avoid the edge effect of the phantom material. The sphere and background counts were repeated three times and the average values were recorded, and then the contrast value for this average was calculated. The contrast value is calculated for each factor (matrix size & motion type) using the same formula described in the previous test. A comparison of the calculated contrast values for each radius was done.

3. RESULTS & DISCUSSION

3.1. 1st Parameter (Matrix Size):

3.1.1. By FBP Reconstruction Method

FBP		Count Rate						Contrast			Av.
		R1		R2		R3					
		L	B	L	B	L	B	R1	R2	R3	
Matrix	64*64	5489.4	204	5038.7	127.7	5103.6	167.9	0.928338	0.950565	0.936299	0.938400744
	128*128	995.4	26.3	950	28	980.3	25.1	0.948517	0.94274	0.95007	0.947109029

FBP		Standard	Contrast (mean)	SD(±)	P Value
Matrix	64*64	1	0.938400744	0.00919508	0.057
	128*128		0.947109029	0.00315351	0.055
P ≤0.05 is defined to be significant					

Table (1): Matrix Size as a function in Contrast values with FBP reconstruction method

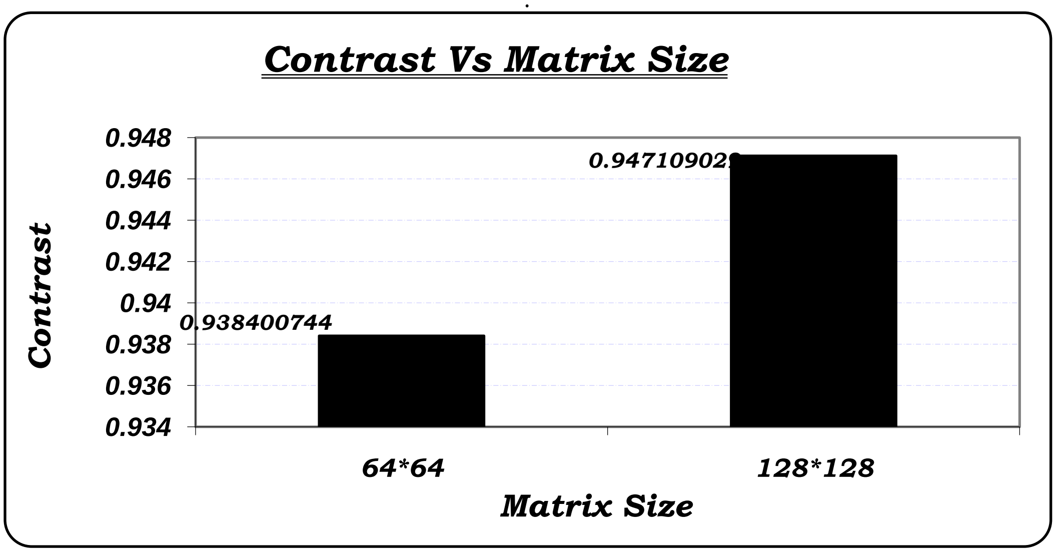

Figure (1): The relation between contrast and matrix size with FBP Reconstruction method.

From the previous data,

- A non-significant difference observed between contrast values in 128*128 and 64*64 matrix size.
- The count rate with 64*64 higher greater than count rate with 128*128 matrix size and that is because the pixel size in 64*64 is larger than pixel size in 128*128 matrix size, at which it is able to collect more counts.

3.1.2. By Iteration Reconstruction Method

Iterative		Count Rate						Contrast			Av.
		R1		R2		R3					
Matrix		L	B	L	B	L	B	R1	R2	R3	
	64*64	4366.5	148.1	4173.3	129.1	4408.8	105.5	0.934391	0.939987	0.95326	0.942545756
	128*128	676.4	19.7	667.1	16.1	654.9	15.3	0.943399	0.952869	0.954342	0.950203259

Iterative		Standard	Contrast (mean)	SD(±)	P Value
Matrix	64*64	1	0.942545756	0.00791287	0.052
	128*128		0.950203259	0.004848823	0.053
P ≤0.05 is defined to be significant					

Table (2): Matrix Size as a function in Contrast values with Iterative reconstruction method.

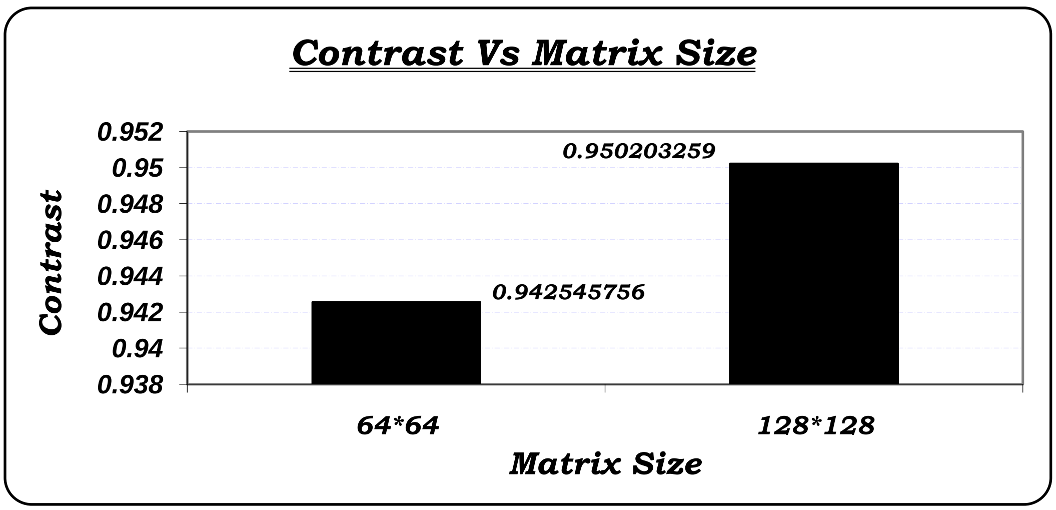

Figure (2): The relation between contrast and matrix size with iterative Reconstruction method.

From the previous data,

- The same effect as in FBP, a non-significant difference observed in contrast values in 128*128 and 64*64 matrix size.
- Count rate with 64*64 is greater than 128*128
- But, generally, the Contrast by using Iterative method is higher than it by using FBP in each matrix size.

The results obtained showed a great difference between count rate taken by matrix 64×64 and that taken by matrix 128×128. This difference concentrated around the image resolution and not affects the contrast. A smaller pixels dimension of 128×128 could display more image detail than the 64×64. There was sever loose of the resolution in the 64 × 64 images, although maximum profiles (opposite to resolution) were larger than 128×128. This could be explained due to the fact that the chosen pixel (matrix) should be less than 1/3 of the expected FWHM .This is matched with the matrix of (128×128).the same results were found by *Farag H et al, 2003*[(6)].

These results are also consistent with the study of the effect of different matrices using clinical bone image as a test pattern which showed that smaller pixels caused improvement in the resolution of images and not affects the contrast. *(Mark F at el 1998)*[(7)].

3.2. 2nd Parameter (Motion Type):

3.2.1 By FBP Reconstruction Method

FBP		Count Rate						Contrast			Av.
		R1		R2		R3					
Motion Type	Contin.	1019.3	51.2	1010.5	50.5	940.9	54.5	0.904344	0.904807	0.890496	0.899882278
	S & S	1024.5	46.2	1004.9	36.9	985.3	32.5	0.913701	0.929161	0.936137	0.92633305
	S&S, Con	995.4	25.3	1012.4	26.5	1080.3	27.1	0.950426	0.948985	0.951057	0.950155737

FBP		Standard	Contrast (mean)	SD(±)	P Value
Motion Type	Contin.	1	0.899882278	0.00663959	0.06
	S & S		0.92633305	0.00937499	0.033
	S&S, Con.		0.950155737	0.00086725	0.012
P ≤0.05 is defined to be significant					

Table (3): Motion Type as a function in Contrast values with FBP reconstruction method.

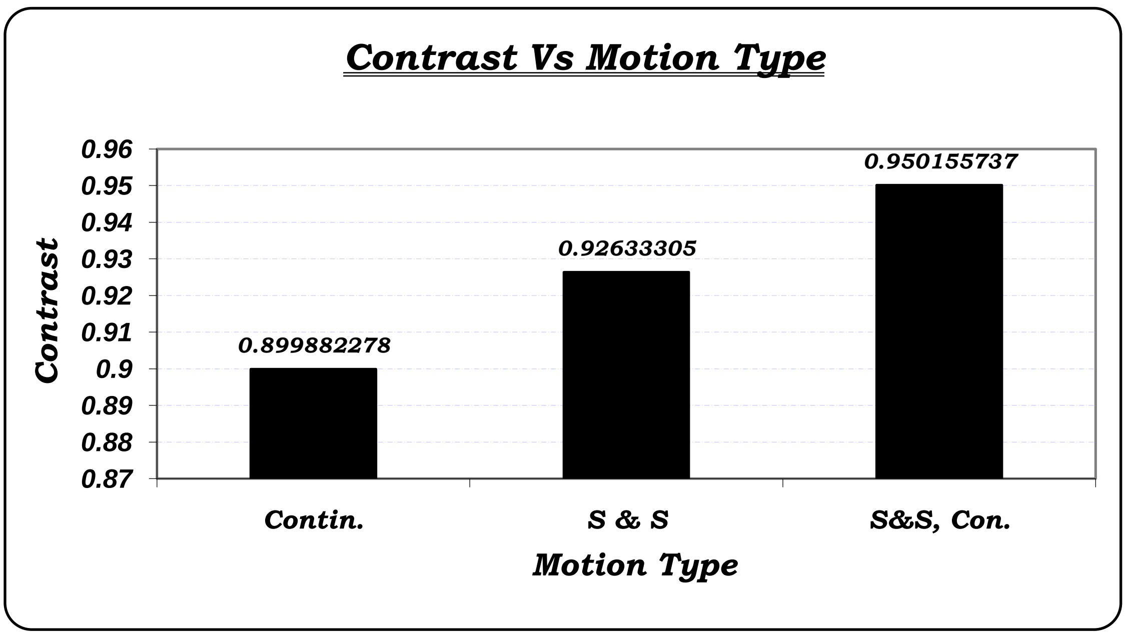

Figure (3): The relation between contrast and motion type with FBP Reconstruction method.

From the previous data,

- A statistically significant difference observed between contrast values by changing motion type.
- S&S continues, gives the best contrast value because it doesn't lose any counts.

2.2.2. By Iteration Reconstruction Method

Iterative		Count Rate R1		R2		R3		Contrast			Av.
Motion Type	Contin.	658.9	38.1	630.6	39.2	618	38.9	0.890674	0.88295	0.881565	0.885063126
	S & S	676.1	27.6	645.2	24.6	659.4	23.9	0.921557	0.926545	0.930045	0.926049362
	S&S, Con	688.5	17.1	679.2	14.2	686.5	15.3	0.951531	0.959042	0.956398	0.955656949

Iterative		Standard	Contrast (mean)	SD(±)	P Value
Motion Type	Contin.	1	0.885063126	0.00400781	0.058
	S & S		0.926049362	0.00348286	0.011
	S&S, Con.		0.955656949	0.003111101	0.009
P ≤0.05 is defined to be significant					

Table (4): Motion Type as a function in Contrast values with FBP reconstruction method.

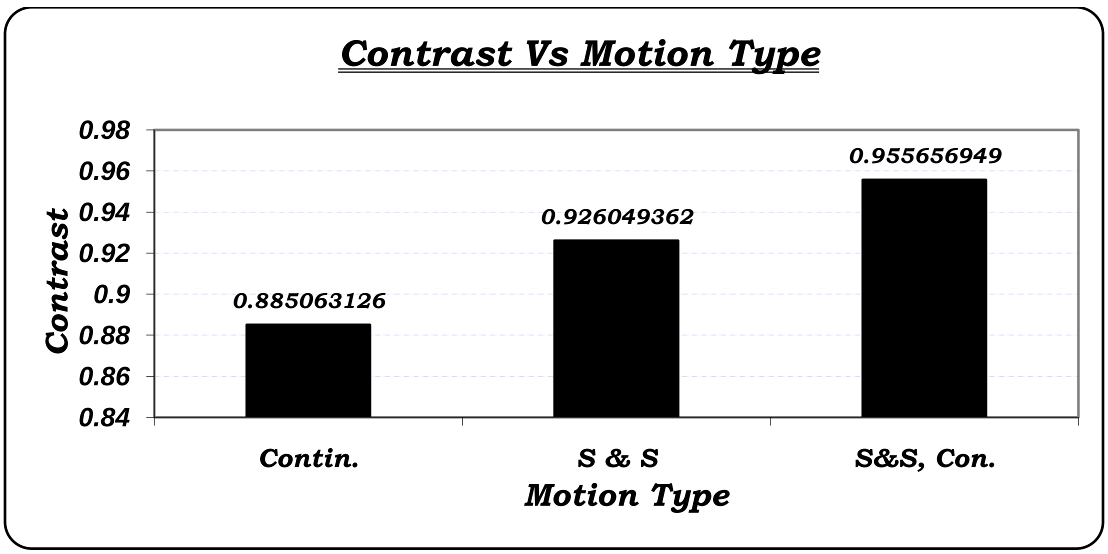

Figure (4): The relation between contrast and motion type with Iterative Reconstruction method.

From the previous data,

- The same effect as in FBP, a statistically significant difference observed in contrast values by changing motion type.
- But, generally, the Contrast by using Iterative method is higher than it by using FBP in each motion type.

In present work, low contrast in addition to blurred lines were observed due to slow motion of gamma camera when it taken count from the object (by continuous motion mode). By using the other two motion types, these blurred lines disappeared and contrast can be increased, due to stop of gamma camera at each projection (view).
When *Tsai.y 2008* proposed a new dynamic SPECT quantification method for estimating the time-varying activity in conventional rotating camera systems with a step & shot protocol, the result showed that the proposed new method can recover the time-activity curve (TAC) with good quality of image [(8)]. In our present study we found that, Iterative algorithm is a preferred alternate method for performing SPECT reconstruction, and over the past few years there has been a shift from filtered back projection to iterative reconstruction in most clinics *(Defrise M et al, 2006.,*

Qi J et 2006) [9] [10] *and (Huang Q et al, 2005,. Tang Q et al, 2005)* [11] [12]. Our results are in line with the improved image quality obtained with iterative reconstruction in a study by *Van Laere et al. (Van Laere K, 2000)* [13].

4. CONCLUSION

It was found that the image contrast by using Iterative method is higher than it by using FBP in most of factors. The big advantage of the iterative approach is that accurate corrections can be made for all physical properties of the imaging system and the transport of γ -rays that can be mathematically modeled. In the present study it was found that, matrix size doesn't affect contrast values in SPECT studies, but contrast by using matrix 128*128 is small higher than 64*64. And S&S continues gives the best contrast value in motion types. In Matrix Size, a non-significant difference observed between contrast values in 128*128 and 64*64 matrix size. The count rate with 64*64 higher greater than count rate with 128*128 matrix size and that is because the pixel size in 64*64 is larger than pixel size in 128*128 matrix size, at which it is able to collect more counts.

Studying the motion Type, There was a statistically significant difference observed between contrast values by changing motion type. S&S continues, gives the best contrast value because it doesn't lose any counts. Finally, contrast of the images in SPECT studies by using Iterative method is higher than it by using FBP in most of physical factors.

5. REFERENCES

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