PHYSICS, ORIGINAL ARTICLE

Contrast Evaluation for a Dual Head SPECT System with Different Energy Peaking-Drift

Saad, I.[*], Mattar, E. [**], Ashour, A.[***]

[*]Department of Clinical oncology and Nuclear Medicine, Faculty of Medicine, CairUniversity-Egypt, [**] Depatrment of radiological sciences,Faculty of applied medical sciences, King Saud University- KSA. , [***]Critical Care unit, Kasr Alini University hospital- Egypt

Abstract

Introduction: Gamma cameras contain energy discriminators that allow only those photons within a specified energy range to be recorded. The process of setting this energy window is called "peaking" the camera. A spontaneous shift in the peak of one head of a triple-head gamma camera was reported to cause an artifact that mimicked ischemia on 201Tl-chloride myocardial SPECT. The characteristic energy distribution results from the collection of total photon energy absorbed by the detector. Physically peaked detectors will center the energy window on the photo peak of selected isotope. **Aim of the study:** Our aim is to evaluate the effect of energy peak drifting for both imaging heads on the image contrast in SPECT. **Materials and methods:** Using a dual head gamma camera, the SPECT phantom with its inserts was scanned using a fixed acquisition protocol and image reconstruction parameters but with changing the energy peak window adjustment which was operated for both heads at different values starting from 120 KeV up to 160 KeV with incremental step of 5 KeV, and the window width was adjusted to be 15% as used in the clinical applications**.**

Results: Our results shows a fluctuation in the image contrast, as there was a gradual increase in the contrast value from 120 KeV setting up to 140 KeV by 31 % and also there was a gradual decrease in the contrast value from 140 KeV setting up to 160 KeV setting by 15 %. The clinical verification for these results was done on cardiac patient who underwent a cardiac scan twice with the normal gamma camera peak setting and then with off-peak setting at 120 KeV (which gives the lowest contrast value for the phantom study). And the case interpretation results confirm the phantom results as there was a significant change in the uptake in the left ventricle wall, with also a verification from the quantitative results by the Quantitave comparison using the quantitative gated SPECT data, that shows a 9% difference in the EF calculated value from the normal to the off-peak setting, which may affect the interpretation and in consequence the further treatment of the patient. **Conclusion**: The energy window peaking for the gamma camera has a proven effect on the planar images, and this was explained and proved in our study to have a significant effect on the contrast of the SPECT images which will affect the interpretation of the clinical results.

Correspondence author:

Dr. Ibrahim Saad

Email: i_elsayed@hotmail.com

Introduction

The purpose of routine quality control is to detect changes in performance from a baseline condition. Typically, baseline characterization is performed by acceptance testing. Thus, routine quality control measurements focus on the detection of a change from baseline, rather than on absolute characterization. [1]
Performance requirements for a scintillation camera based SPECT system are more stringent than the requirements for planar imaging with the same camera.
A routine quality control program must include a sufficiently comprehensive suite of individual measurements to ensure adequate sensitivity to detection of detrimental changes in performance. At the same time, the criteria used to judge the outcome of routine quality control must not be so strict as to misleadingly identify insignificant changes as important.[2]

The single-head gamma camera SPECT system is formed of one detector. The single-head camera SPECT systems are inexpensive compared to multihead SPECT systems, and its quality control is fairly straight-forward.

The disadvantages of single-head SPECT systems include relatively low sensitivity compared with multihead systems and thus, a generally longer patient acquisition time. With single-head SPECT systems, scan times are seldom less than 15–20 min and are frequently 30 min or more. Typically, acquisition times of more than 30 min cause significant patient actually provide scans inferior to those done using a shorter acquisition-time-scan, due to a higher degree of patient motion. [3, 4]
In case of multi-head systems, they are formed of two or three detectors. It is designed specifically for high throughput; high performance single photon emission computed tomography (SPECT) imaging .The system can perform both circular orbit and non circular-orbit capability. In addition other study types such as planer dynamic, gated and gated SPECT are accommodated. [5]
Camera systems with two or more heads surround the patient with more detectors and offer more optimal spatial Resolution/sensitivity characteristics than are available with a single-head system. These combinations assume that the data from different heads are matched in gain, orientation and offset, so that the data can be combined. The quality control procedures must ensure that adding the data from different heads can be performed without artifact [6].

For multihead 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:

$$\textbf{Contrast} = \frac{(\text{C sph.} - \text{C bgd})}{(\text{C sph.} + \text{C bgd})} \qquad \text{Equ (1)}$$

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 delectability of small lesions.

It is affected by many different properties of the system, in particular energy resolution, the contribution of scatter and the reconstruction filter. Tomographic contrast decreases as the size of the object becomes comparable to or smaller than, the spatial resolution of the system, or when the object only partially fills the reconstruction slice. These two effects are called the partial slice filling effect and partial volume effect respectively. [7]

Gamma cameras contain energy discriminators that allow only those photons within a specified energy range to be recorded. The process of setting this energy window is called “Peaking” the camera. [8]

A spontaneous shift in the peak of one head of a triple-head gamma camera was reported to cause an artifact that mimicked ischemia on 201Tl-chloride myocardial SPECT [9].

The characteristic energy distribution results from the collection of total photon energy absorbed by the detector. Physically peaked detectors will center the energy window on the photo peak of selected isotope. [10]

Aim of the study

The aim of the study is to evaluate the effect of energy peak drifting for both imaging heads on the image contrast in SPECT.

Materials and methods

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.

Total Performance Phantom:

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.

Technetium Generator:

All images were acquired with the use of radioisotope 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.

Methods:

Using of the (Axis) Dual head gamma camera, with Low Energy high resolution collimators (LEHR), the SPECT phantom with its inserts was scanned using the following fixed acquisition protocol and image reconstruction parameters in the following Tables [1] and [2] respectively, the only variable was the energy peak window shift which was adjusted for both heads at different values starting from 120 KeV up to 160 KeV with incremental step of 5 KeV, and the window width was adjusted to be 15% as used in the clinical applications.

Table (1) : Acquisition parameters

Acquisition parameter	Camera information Acquisition
Matrix size	64*64
Angle of rotation	360^o
Starting position for detector1	0^0
Zooming	1
Pixel size	6.4mm
Collimator	LEHR
Phantom distance(CD)	10cm
Time per view	25 sec
Number of view	64
COR correction	Applied
Camera orbit	Circular
Energy window width	15 %

Table (2): Show processing parameters

no.	Processing Parameters	Camera information
1	Flood correction	Applied
2	Attenuation correction	Applied (auto boundary)
3	Back projection filters	Butterworth (Cut off 0.6-Order 7)

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The SPECT phantom filled with water mixed with (20 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 the center of the peak, those images were reconstructed with the reconstruction parameters listed in table (2) 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.

Verification for the results was done on cardiac patient who underwent a cardiac scan with the normal gamma camera acquisition setting using 140 KeV adjustment of the energy peak and also underwent another scan using 120 KeV adjustment for the energy peak for both heads, quantitative and qualitative comparison between images was done to verify the results of the phantom testing.

Results

The results shown in **figure (1)** which describe the relationship between the sphere count and the change in the peak energy indicate that there is a variation in the sphere counts with changing the peak energy. This variation is described as follows, there was an increase in the sphere count by 38% from 120 KeV up to 135 KeV at which the maximum sphere count was recorded. There was a sharp decrease in the sphere count by 94.5 % from 135 KeV up to 160 KeV.

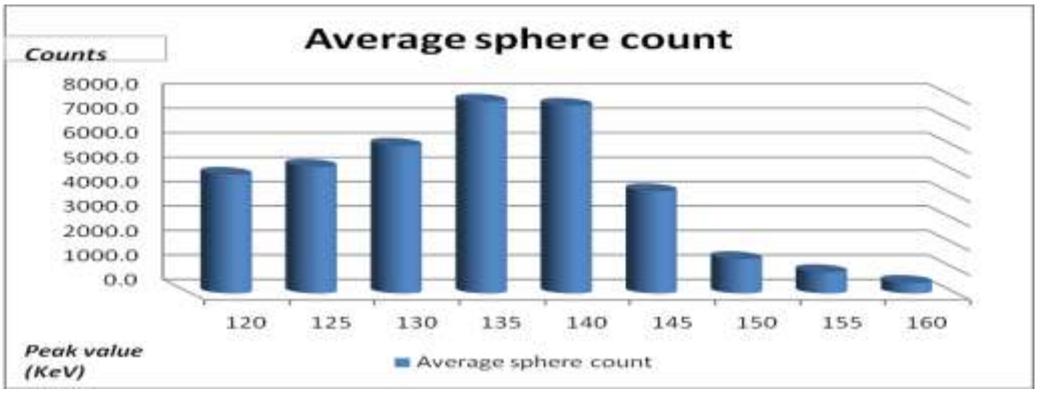

Figure (1): Relationship between the changes in the peak energy with the average counts in the measured sphere.

The results shown in figure (2) which describe the relationship between the changes in the peak energy with the average background counts in the measured sphere indicates that there was a gradual decrease in the sphere background count from 120 KeV up to 135 KeV by 34.5 %. Then it was followed by a sharp decrease in the sphere background counts by 76.4 % from 135 KeV up 160 KeV.

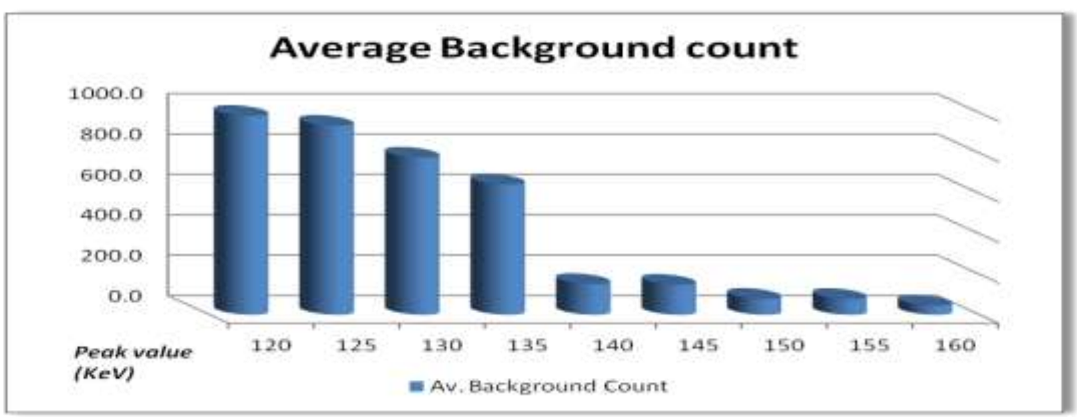

Figure (2): Relationship between the changes in the peak energy with the average background counts for the measured sphere with the same size.

From the results illustrated in figure (3), which describe the relationship between the changes in the peak energy with the average contrast value for the measured sphere, and it indicates that the maximum contrast value was recorded for the 140 KeV as it is the normal adjustment for the gamma camera setting, and the lowest contrast value was recorded for the 120 KeV setting. There was a gradual increase in the contrast value from 120 KeV setting up to 140 KeV by 31 % and then there was recorded a gradual decrease in the contrast value from 140 KeV setting up to 160 KeV setting by 15 %.Table three shows the results for the contrast value for each peak value tested in the current work and show also the standard deviation for the repeated measurement for these contrast values. From the table it was clearly shown that the best contrast value was found at 140 KeV measurements to be (0.96 ± 0.01), and the lowest contrast value was found at 120 KeV measurements to be (0.66± 0.02).

Table (3): The contrast value for the three readings with their standard deviation as well as the average contrast for each peak value.

Peak Value	Contrast			SD	Average
	R 1	***R 2***	***R 3***		
120	**0.66**	**0.64**	**0.68**	**0.02**	**0.66**
125	**0.69**	**0.69**	**0.70**	**0.00**	**0.69**
130	**0.79**	**0.79**	**0.73**	**0.04**	**0.77**
135	**0.84**	**0.86**	**0.84**	**0.01**	**0.85**
140	**0.97**	**0.96**	**0.95**	**0.01**	**0.96**
145	**0.94**	**0.93**	**0.93**	**0.01**	**0.93**
150	**0.92**	**0.87**	**0.89**	**0.03**	**0.89**
155	**0.85**	**0.79**	**0.86**	**0.03**	**0.83**
160	**0.83**	**0.82**	**0.78**	**0.03**	**0.81**

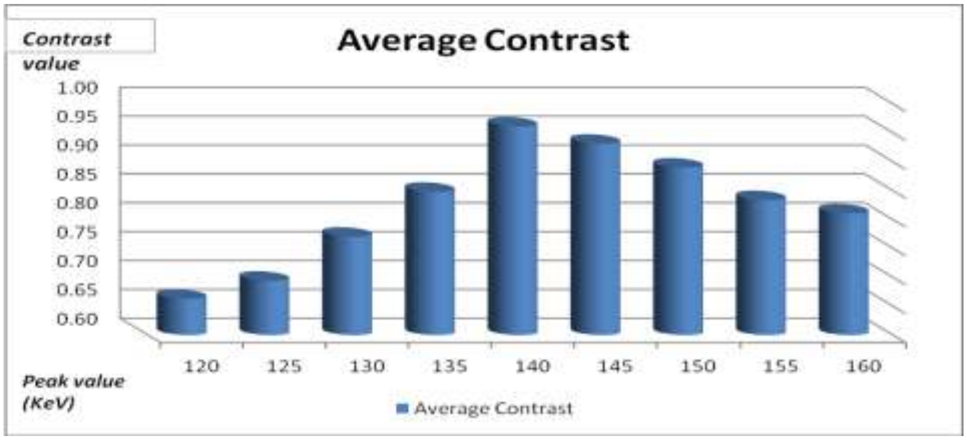

Figure (3): Relationship between the changes in the peak energy with the average contrast value for the measured sphere.

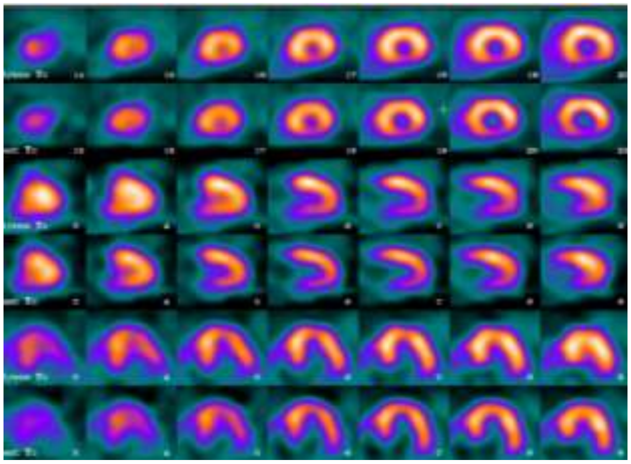

Figure (4): A cardiac patient scan using the normal window peaking setting at 140 KeV for both imaging heads.

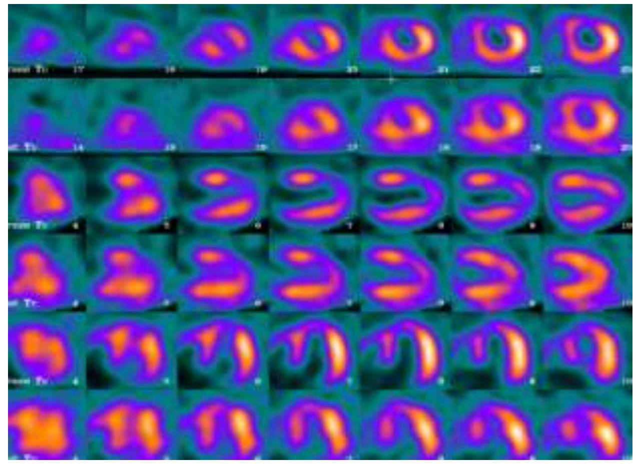

Figure (5): Images for the same patient using the drifting for the window peaking to 120 KeV for both imaging heads.

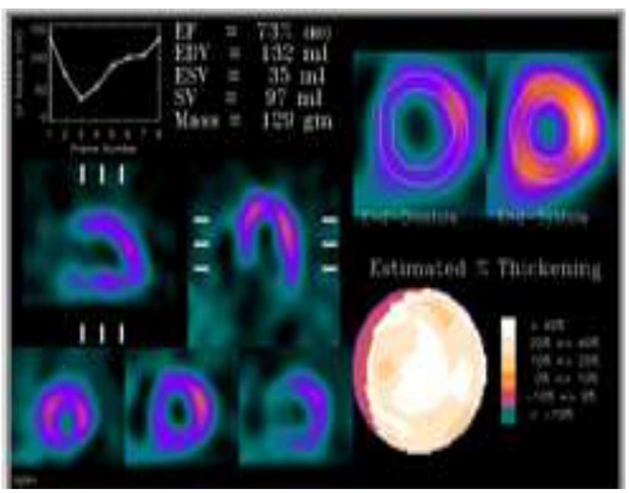

Figure (6): Quantitative results for a cardiac patient scan using the normal window peaking setting at 140 KeV for both imaging heads.

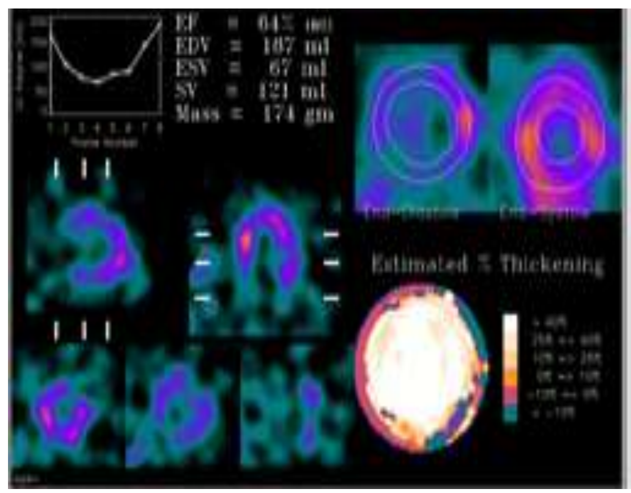

Figure (7): Quantitative results for the same patient using the drifting for the window peaking to 120 KeV for both imaging heads.

Figures 4 and 5 shows qualitative comparison for the same cardiac patient for 140 KeV and 120 KeV window setting, from which it is clear that the contrast of the image is highly acceptable and there was a better edge detection for the Lt. ventricle wall for the 140 KeV setting rather than that of 120 KeV setting which verify the results of the phantom study. Also there was a significant change in the uptake in the left ventricle wall which may cause false interpretation for the case. These results were assured by the quantitative comparison for the results of the same patients shown in figures 6 and 7 for 140 KeV and 120 KeV setting respectively using Quantative Gated Spect(QGS) software provided with processing computer of the gamma camera.These results show a change in the Ejection Fraction of the patients from 73 % for 140 KeV setting to 64% for 120 KeV setting with 9% difference in the calculated value which may affect the treatment of the patient.

Discussions

Gamma cameras contain energy discriminators that allow only those photons within a specified energy range to be recorded. The process of setting this energy window is called "peaking" the camera. [8] A spontaneous shift in the peak of one head of a triple-head gamma camera was reported to cause an artifact that mimicked ischemia on 201Tl-chloride myocardial SPECT [9]. The characteristic energy distribution results from the collection of total photon energy absorbed by the detector. Physically peaked detectors will center the energy window on the photo peak of selected isotope. [10]

S.H.A. Al-Lehyani [11] studied the flood images acquired with different off-peak shifts. The results showed differences in the images quality which could be due to the off-peak shift. He concluded that in planer and SPECT images, the uniformity and quality of images are affected by a change of the off-peak shift and this effect becomes more prominent at the peak shift of more than 2%. Thus, it was recommended that gamma camera used for this purpose should be operated with off-peak shifts less than 2% otherwise images with bad quality and uniformity will lead to false diagnosis. In that study he studied a small range of peak shift and its effect on the uniformity in the first place and image quality which is a broad term that must be specified in a quantitative parameter. While in our study the range of energy peak drifting was done on a range from 120 KeVup to 160 KeV, with the same clinically applied window width which is 15%,and the studied parameter was the contrast which is a quantitative parameter in SPECT imaging used to evaluate the image quality.The results for the sphere count indicate that there is a variation in the sphere counts with changing the peak energy. This variation was described as follows, there was an increase in the sphere count by 38% from 120 KeV up to 135 KeV at which the maximum sphere count was recorded, and then there was a sharp decrease in the sphere count by 94.5 % from 135 KeV up to 160 KeV. This variation in the sphere count is almost matching the variation of count rate of the technetium (Tc-99m) energy spectrum, which explains that change in the sphere count.This is also can explain the variation of the sphere background count that gives a high background count at the area of scatter in the technetium (Tc-99m) energy spectrum, and also explain the decrease in the background counts after 135 KeV by 76.4% to that of 160 KeV energy peak.As regard to the results for the contrast there was a gradual increase in the contrast value from 120 KeV setting up to 140 KeV by 31 %, this increase is due to the decrease in the background count when it moves from the area of the scatter in the Tc-99m energy spectrum. Also there was a recorded gradual decrease in the contrast value from 140 KeV setting up to 160 KeV setting by 15 %, but the visualization of the sphere was not good as the quantitative results shows because there was a decrease in both the sphere and background count as well which in turn decreases the contrast with small percentage (15%).As regards to the clinical verification of the phantom results, Wei-Jen Shih et.al, [9] have studied Four patients whom underwent imaging on a dual-head gamma camera on the same morning with an off-peak status of 1 head of the dual-head camera occurred with 99mTc-labeled compounds and resulted in artifacts on myocardial gated SPECT images. The off-peak status was caused by a malfunction of a photomultiplier tube. And they found that degradation of planar images, such as bone scans, because of off-peak status appears to be easily identifiable, but for gated cardiac SPECT findings resulting from off-peak status, including reversible defects, left ventricular wall motion abnormalities, and faulty LVEF, were not easy to discover. Also Momennezhad et al.,[(12)] have studied a case of spontaneous photopeak shift during acquisition of a dynamic renal scan, and they found that from the beginnings of frame number 24, the count in following frames are severely decreased, resulting in complete loss of contrast in some of the frames. And from the quantitative analysis they had, the total counts were fluctuating in different frames. These fluctuations in count density suggested an unstable system, which was due to malfunctioning hardware that was responsible for transient shift of photopeak to lower portion of the energy window and beyond it. With that transient shift a static imaging may not be affected if predefined count was set for acquisition. However the required time for acquisition may be increased. On the other hand if predefined time was set for acquisition, the image count will be significantly decreased.In the current study the same concept was assured because of the change in the contrast of the cardiac images from 140 KeV setting to that of 120 KeV setting there was a significant change in the uptake in the left ventricle wall which may cause false interpretation for the case. And this was also confirmed by the Quantitave comparison of the quantitative gated SPECT data that shows a change in the Ejection Fraction (EF) of the patients from 73 % for 140 KeV setting to 64% for 120 KeV setting with 9% difference in the calculated value which may affect the interpretation and in consequence the further treatment of the patient.

Conclusion

The energy window peaking for the gamma camera has a proven effect on the planar images, and this was confirmed in our study to have a significant effect on the contrast of the SPECT images which will affect the interpretation of the clinical results. Using a dual head gamma camera, the SPECT phantom with its inserts was scanned using a fixed acquisition protocol and image reconstruction parameters but with changing the energy peak window adjustment which was operated for both heads at different values starting from 120 KeV up to 160 KeV with incremental step of 5 KeV, and the window width was adjusted to be 15% as used in the clinical applications. The results shows a fluctuation in the image contrast, as there was a gradual increase in the contrast value from 120 KeV setting up to 140 KeV by 31 % and also there was a gradual decrease in the contrast value from 140 KeV setting up to 160 KeV setting by 15 %. The clinical verification for these results was done on cardiac patient who underwent a cardiac scan twice with the normal gamma camera peak setting and then with off-peak setting at 120 KeV (which gives the lowest contrast value for the phantom study). And the case interpretation results confirm the phantom results as there was a significant change in the uptake in the lt. ventricle wall, with also a verification from the quantitative results by the Quantitave comparison using the quantitative gated SPECT data, that shows a 9% difference in the EF calculated value from the normal to the off-peak setting, which may affect the interpretation and in consequence the further treatment of the patient.

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