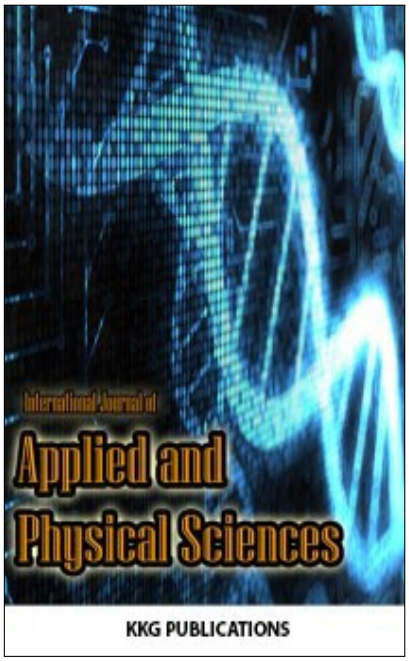

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PHANTOM STUDY: NON UNIFORMITY QUANTITY OF TECHNETIUM-99M IN DIFFERENT SEGMENTS OF MYOCARDIAL SPECT IMAGE

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Abstract. It is important to quantify the uniformity of pixel intensity in myocardial SPECT images at varied tomographic slices. The distribution of pixel densities in SPECT images mirrored the distribution of radionuclide tracer within target organs. Non-uniformity of pixel intensities in SPECT images may lead to an incorrect diagnosis of myocardial infarction. Myocardial phantom positioning is one of the many reasons that affect pixel intensity distribution. In this research work, the Myocardial fabricated phantoms were used to mimic the myocardial wall at end-diastole and end-systole. These phantoms were separately placed at four different positions on the imaging table. The purpose of this study was to quantitatively determine the pixel intensity distribution of the reconstructed myocardial SPECT images and its relation to phantom positioning on the imaging table. Results show non-uniformity of pixel intensity of SPECT short-axis image whether the myocardial phantom was end-diastole or end-systole. The non-uniformity was not significant when reconstructed SPECT images of both stages were compared. This non-uniformity became significant ($p<0.05$, sigma plot) with an increasing position of both myocardial phantom stages. In addition, non-uniformity became in big degree in an inferior segment more than a segment in septal, anterior, and lateral particularly, when the myocardial phantoms were positioned at 15 cm off-center.

INTRODUCTION

Single Photon Emission Computed Tomoghraphy (SPECT) has been routinely used to detect myocardial infarction in myocardial perfusion imaging [1]. Gamma camera is used in SPECT imaging to obtain images from different angles around patient on an imaging table injected by isotopes [2]. For SPECT image to be of perfect quality, camera heads must work in best conditions to avoid non-uniformity of isotopes distribution within organs [3].

Variations of energy and uniformity of tomography reconstruction need more calibration for center of rotation [4-6]. Center of rotation error is one of the common reasons of artifact causing non-uniformity of SPECT image [7]. The patient position should be convenient with the imaging table to correspond with the infected organ, below the rotating gamma camera heads [8].

Many studies in nuclear SPECT imaging focused on appropriate patient body positioning on an imaging table with the gamma camera rotation [9-11]. The positioning of the cardiac through gamma camera orbit is important, and that an off centre position may make artifacts [12]. In myocardial perfusion infraction (MPI), partial volume considered is one of the factors which affects the image quality causing inhomogeneity of SPECT image, and therefore afects diagnose accuracy. Many investigations reported that the inhomogeneity of images is due, at least in part, to physiologic variations in uptake around the myocardium [13, 14]. Inhomogeneity of myocardial wall activity can be created by partial volume related effect in SPECT [15]. The regional maximum counts per pixel in the image have no longer relationship with concentration of radionuclide at that region, but will vary with thickness and the size of the object [16].

SPECT with technetium-99m (^{99m}Tc) has used parameters to get optimized imaging. However, nuclear medicine imaging has not attained the level of standardization, due to a big number of physical parameters affecting image uniformity. The purpose of this study was to determine quantified non-uniformity of pixel density in stage (end diastole, end systole), as well as effect of the positioning of phantom on uniformity of the pixel intensity in myocardial SPECT images.

METHOD AND MATERIAL

Two fabricated phantoms of plastic were used for this study. $Z_{eff} = 0.57$, density = 0.93 g/cm^3) [17]. The fabricated phantoms are used to simulate the human tissue of the myocardial wall of left ventricle. Measurements of the fabricated phantoms of two chambers (inner, outer) were designed to mimic left ventricle in both stages (end diastole and end systole) (fig. 1).

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The inner chamber of the phantom was filled with water while the space between two chambers was filled with a thoroughly mixed Technetium-99m solution (0.43, 0.31)(mci) for end diastole and end systole respectively [18].

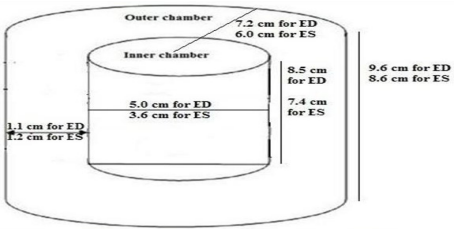

Fig. 1 . Schematic of myocardial phantoms to simulate myocardial wall of left ventricle for two stages,end diastole and end systole

Two phantoms were scanned on an imaging table at four positions: the centre, 5, 10, and 15 (cm) to mimic the range of normal cardiac position (fig. 2).

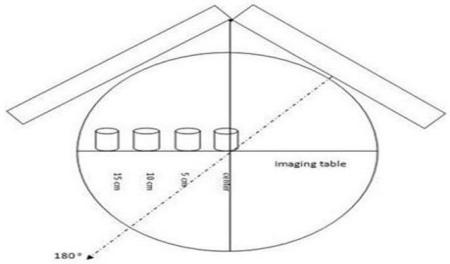

Fig. 2 . Schematic of 4 phantoms in straight orientation on an imaging table

A dual-head gamma camera (Discovery NM/CT670 Pro) with parallel hole collimator was used. ^{99m}Tc 140 keV at 20% energy window was used in this research work. A 180° circular orbit from the right anterior to the left inferior was applied throughout all acquisitions. The total acquisition time is 12.5 minutes. The matrix size is 64×64, with a pixel size of 6.8×6.8 mm. Butterworth filter (order 10, 0.4 Frequency) was applied in the reconstruction. SPECT images were quantified using image J 1.48v. Using the software the short-axis shape is divided into 8 segments (fig 3). Pixel intensity profiles at the mid regions in all the 8 segments were then determined by mean pixel values in all 8 segments representing the uniformity of ^{99m}Tc tracer (recorded as a database). Database was obtained at each position of the phantom on an imaging table for both phantom configurations.

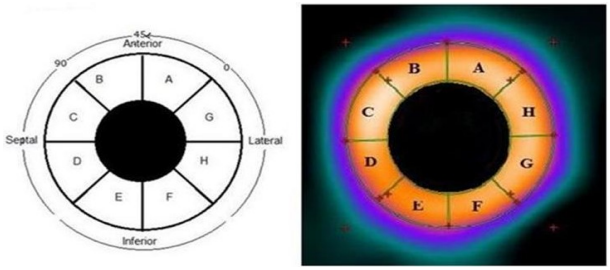

Fig. 3 . Myocardial segmentation of short-axis view. 8 segments (A-H) are used for mid-ventricle. A pair of segments (90° segments) symmetry one region of myocardial wall, where (A,B) anterior, (C,D) septal, (E,F) inferior, and (G,H) lateral

Theoretically, the variability of pixels distribution in SPECT image should be 0%. The distribution of pixel intensity in the reconstructed SPECT images was quantitatively analyzed as the percentage of non-uniformity as follows:

$$nonuniformity(\%) = \frac{maxvalue - minvalue}{maxvalue + minvalue}$$

The mean percentage of non-uniformity was calculated by averaging the percentage non-uniformity of 3 repeated SPECT acquisitions-test, sigma plot was used for two groups' comparison for end diastole and end systole image acquisitions. Probability values at <0.05 were considered statistically significant.

RESULTS AND DISCUSSION

Although the phantoms at both stages (end diastole and end systole) were filled with uniform (homogeneous) solutions of ^{99m}Tc, the distribution of pixel intensities of the reconstructed SPECT image was not uniform (fig 4). The non-uniformity may be caused by many factors such as photon scattering and self-attenuation, undetected photons of photo peak window, imaging table, and positioning of the phantom itself.

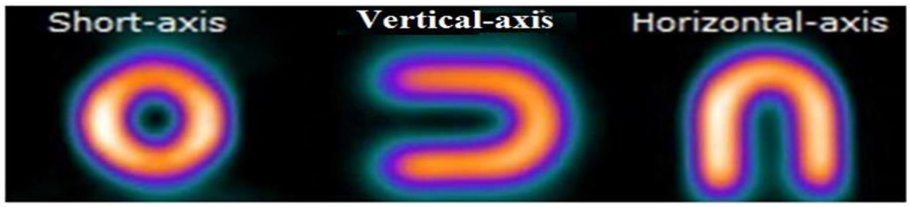

Fig. 4 . Non-uniformity of SPECT image in 3 views; short-axis, vertical-axis, and horizontal-axis

At one glance the distribution of ^{99m}Tc appeared as uniform in the reconstructed SPECT images. However, the quantitative analysis study shows that the pixel intensities profiles were infact not uniform (table 1 and table 2). The non-uniformity in reconstructed SPECT image for end-diastole was slightly greater than in end systole. This non-uniformity was expressed as mean non-uniformity of the acquired SPECT images, and was not significant when the quantitative analysis results of the end diastole and end systole were compared. Similarly, the reconstructed SPECT slices also showed no uniformity when the normal myocardial phantom was positioned off-centre as shown in (fig. 5 and fig. 6).

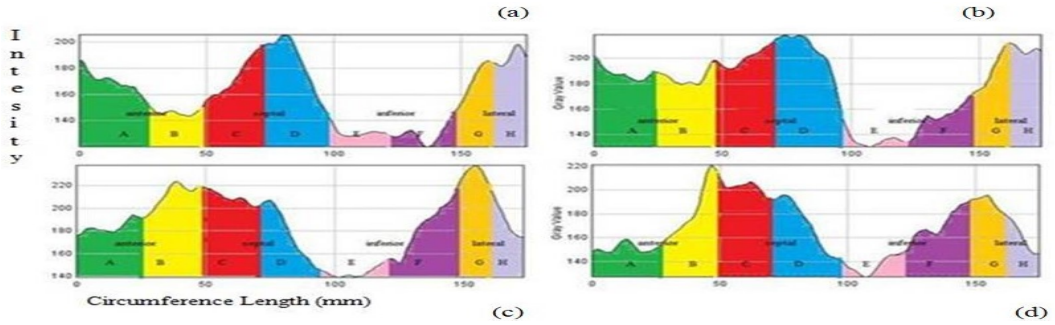

Fig. 5 . Demonstration of non-uniformity in 8 segments

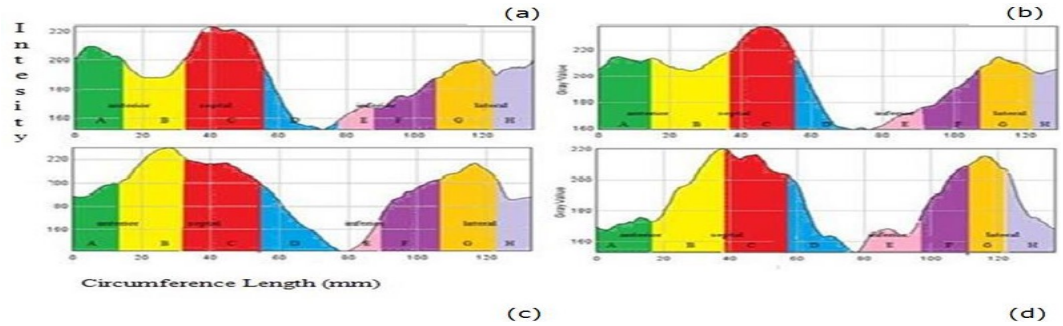

Fig. 6 . Demonstration of non-uniformity in 8 segments

TABLE 1
NON-UNIFORM INTENSITIES IN 8 SEGMENTS

	Anterior		Septal		Inferior		Lateral	
Position	A	B	C	D	E	F	G	H
center	5.5	5.2	3.4	3.8	6.1	6.8	3.5	3.9
5	6.8	6.8	4.3	4.8	6.8	7.1	3.9	4.3
10	8.4	8.3	5.8	6.0	6.7	7.4	5.9	6.2
15	10.2	10.3	7.0	7.3	7.0	7.7	7.4	7.6

TABLE 2
NON-UNIFORM INTENSITIES IN 8 SEGMENTS

	Anterior		Septal		Inferior		Lateral	
Position	A	B	C	D	E	F	G	H
center	5.4	5.0	3.4	3.5	5.9	6.3	3.3	3.8
5	6.7	6.6	4.2	4.5	6.2	7.0	3.6	4.0
10	8.1	8.3	5.5	5.9	5.4	7.3	5.7	5.8
15	10.0	10.3	6.9	7.1	6.7	7.3	7.0	7.5

The non-uniform pixel intensities of the reconstructed SPECT images at end diastole and end systole stages were not significant ($p<0.05$) when compared. However, the non-uniform pixel intensities were significant ($p>0.05$) particularly in the anterior and inferior segments) as the positioning moved.

The non-uniform pixel intensities in all segments of the reconstructed SPECT slices are at the minimum values when the phantoms were positioned at the center in comparison with off-center. On the other hand, non-uniform pixel intensities at either end diastole or end systole became greater in inferior segments when the positioning of the phantom became eccentric until 5 cm off-center, but this non-uniformity changed to be greater in anterior wall when positioning was increased by 10 and 15 cm off-centre. The greatest non-uniformity occurred in the inferior wall due to loss of counts may be caused by artifacts related to poor spatial resolution or non-attenuating gamma rays within the phantom materials. This artifact was observed in clinical studies [19, 20], and in phantom study [21]. Positioning of the phantom may be one of the reasons causing the loss in counts. The effect is clearly seen that the non-uniform pixel intensities changed to be greater in the anterior wall instead of in the inferior wall positioning were increased from 5 cm to 15 cm. The previously elaborated reasons which caused artifacts may incur due to the difference in distribution of pixel intensities (density) in the reconstruction of myocardial SPECT image, therefore degradation of SPECT image quality.

The gamma camera heads are at different distances from the Myocardial phantom. Segments of the left myocardial walls closest to the detector are better resolved. The non-uniformity of the pixel intensities increased significantly ($p>0.05$) as the phantoms were positioned off the centre. The positioning impacts were significant ($p>0.05$) at both end diastole and end systole stages.

Study showed that acquired SPECT images showed significant ($p>0.05$) non-uniform pixel intensities which were observed when a non-defected MI phantom was scanned. The uniformity of SPECT short axis images in base segment was largely convergent between end diastole and end systole configurations, when the myocardial phantom was positioned at different positions on the imaging table. The non-uniformity in myocardial SPECT images was observed significant ($p>0.05$) at the end diastole and end systole stages. The pixel intensities' uniformity of base segment SPECT images was observed improved, as the positioning increased on an imaging table.

CONCLUSION

The non-uniformity of pixel intensity in myocardial SPECT images increases with increasing the positioning on an imaging table and becomes in big degree in an inferior wall, particularly with big eccentric of the positioning. This non-uniformity is not significant when end diastole and end systole SPECT images are compared.

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