INFLUENCE OF SOURCE-TO-COLLIMATOR DISTANCE ON IMAGE QUALITY IN SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY: A PHANTOM STUDY

SAYED INAYATULLAH SHAH, PhD (CORRESPONDING AUTHOR)
DEPARTMENT OF DIAGNOSTIC IMAGING AND RADIOTHERAPY, KULLIYYAH OF ALLIED HEALTH SCIENCES, INTERNATIONAL ISLAMIC UNIVERSITY MALAYSIA, JLN SULTAN AHMAD SHAH, 25200 KUANTAN, PAHANG, MALAYSIA.
inayatullah@iium.edu.my

NUR `AMIRAH BINTI YUSUP
DEPARTMENT OF DIAGNOSTIC IMAGING AND RADIOTHERAPY, KULLIYYAH OF ALLIED HEALTH SCIENCES, INTERNATIONAL ISLAMIC UNIVERSITY MALAYSIA, JLN SULTAN AHMAD SHAH, 25200 KUANTAN, PAHANG, MALAYSIA.

ABSTRACT

Introduction: Source-to-collimator distance is one of the physical factors that can affect the quality of images generated in Single Photon Emission Computed Tomography (SPECT). Therefore, this work investigated the effect of source-to-collimator distance on image quality in terms of uniformity, hot and cold regions contrast and signal-to-noise ratio (SNR). **Methods:** SPECT data were acquired by scanning R.A. Carlson's cylindrical phantom using Philips ADAC Forte dual head gamma camera with parallel-hole low energy high resolution (LEHR) collimator. Five different settings of collimator to phantom distance were applied which were 26, 27, 28, 29 and 30cm. Reconstruction of the images was done by using filtered back-projection (FBP) in conjunction with Butterworth filter with cut off frequency of 0.35 cycle/cm and order 5. The reconstructed images were evaluated quantitatively via standard deviation, contrast and SNR. **Results**: It was found from the quantitative analysis that the standard deviation in the counts of uniform regions increased whereas the contrast and SNR of most hot and cold regions decreased with the increase in phantom-to-collimator distance. **Conclusion**: The farther the object to be imaged from the gamma camera collimator results, the poorer image quality of the reconstructed image. Therefore, in SPECT clinical examinations attention on the source-to-collimator distance needs to be given in order to achieve the diagnostic quality images.

KEYWORDS: SPECT, Image quality, Source-to-Collimator distance, Spatial resolution

INTRODUCTION

In medical imaging, there are several modalities that are used to generate medical diagnostic images and one of them is nuclear medicine imaging. This kind of modality uses radionuclides and is commonly being used for diagnostic and therapeutic purposes as well as biomedical research (International Atomic Energy Agency, 2013). In terms of fundamental concept in nuclear medicine study, it involves the administration of radiolabelled compound which is called as the radiopharmaceutical, tracer or radiotracer into the body. The administered radionuclide decays in the patient's body and emits the radiation, such as gamma rays, with the energy that is sufficient to exit the body without undergoing scatter or attenuation. A gamma camera is positioned externally to detect these gamma rays and the distribution of the radiopharmaceutical throughout the body forms the image accordingly (Cherry et al, 2012).

A number of physical factors can affect the image quality in SPECT imaging (Frank 2019, Noori-Asl 2020). Among them is source-to-collimator distance as the collimated gamma camera varies with it. The source to detector distance becomes among the important factors (Prekeges, 2011) which is the principle limiting factor in spatial resolution. Hence, the spatial resolution becomes poorer as the distance of the source from the collimator increases (Cherry et al, 2012). Typically, at a distance of 4 - 6cm from the collimator, the spatial resolution degrades by a factor of 2. Sorenson and Phelps 1987 (as cited in Li et al, 1998) also mentioned that spatial resolution variation as a function of distance from collimator surface is a major obstacle for quantitative imaging.

In SPECT imaging, the effect of distance-dependent resolution on the blurring will increase with the deeper structures (Glick et al, 1994). Thus, the cardinal rule that can be followed is that the source-to-detector distance must be kept to a minimum by bringing the detector to the patient as close as possible for having an optimal gamma camera imaging. Other than that, in 2006, Motomura and colleagues stated that, in SPECT, spatial resolution is the most important factor in determining image quantification and image contrast. Especially in clinical studies, detection sensitivity for tumour lesions and defect areas greatly depends on spatial resolution. This is true as the improved resolution can lead to higher lesion contrast, and improves the accuracy of lesion detection (Kohli et al, 1998).

In SPECT imaging, since it makes use of the collimated gamma camera that has to rotate around the patient for taking various projections in image generation, the physical factor that could affect the performance is source-to-collimator distance. This is because the source-to-collimator distance could influence the image quality of the SPECT images. In clinical settings, there is no specific patient and collimator distance to be set in the SPECT image acquisition. The only rule that has been applied is that the collimated gamma camera should be placed as close as possible to the patient without affecting the patient's comfort as well as disturbing the rotational movement of the gamma camera.

However, this condition might change according to the patient size. The size of the patient could alter the patient and collimator distance and this could influence the quality of the image. The obese patients will cause the rotational radius of the detector farther in order to allow it to rotate freely. On the other hand, this might be critical in preservation of the image quality since the shape of the human body is not circular but is elliptical. It is because the distances between the source and the collimator at the anterior or posterior aspects of the patient are longer as compared to the lateral aspect. Thus, this study made an attempt to investigate the impact of source-to-collimator distance on image quality.

METHODS

The SPECT camera used in this study was Philips ADAC Forte dual head gamma camera and the type of collimator used was parallel-hole, low energy high resolution (LEHR) collimator. An R.A. Carlson SPECT phantom which consisted of a cylindrical tank with linearity, hot and cold lesion inserts was used. The dimensions of the cylindrical tank were (21.6cm OD x 20.3cm ID x 30.5cm long). The hot lesion insert consisted of eight pairs of holes with different sizes in diameter (4.7, 5.9, 7.3, 9.2, 11.4, 14.3, 17.9 and 22.3 mm) drilled in a "V" shape through an acrylic block. The cold lesions insert consisted of six plastic rods of different sizes of diameter (5.9, 7.3, 9.2, 11.4, 14.3, 17.9mm) along the edge of the insert. The height of the cylindrical tank was sufficient. Even when all the inserts were placed into the tank there was still space left for uniformity measurement as shown in Figure 1.

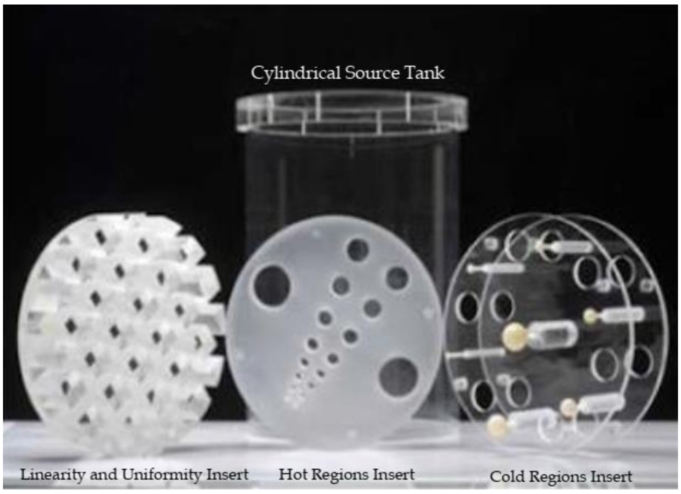

Figure 1 R.A Carlson's SPECT Phantom and inserts (Bailey et al, 2014)

SPECT data acquisition

The R.A. Carlson acrylic cylindrical tomography phantom was prepared by placing the linearity, cold and hot region inserts into the tank. This was followed by filling of water into the tank. The Technetium-99m (Tc-99m) solution with 20.8mCi activity was injected into the water-filled phantom and it was mixed thoroughly. The phantom was then positioned on the table couch in horizontal position and it was centred to the field of view (FOV) of the gamma camera. The initial source-to-collimator distance used was 15.9 cm as it is the distance that allows the gamma camera head to rotate freely without any collision with the table couch as well as the phantom. The procedure was repeated by adjusting the radius of rotation of the gamma camera head to 16.9 cm, followed by 17.9 cm and 18.9 cm. The standard symmetrical energy window was set to 20% (126 to 154 keV) centred at 140 keV and the matrix size used was 128x128x16. One hundred and twenty projections over 360^{o} clockwise rotation were taken and the time for each projection was 20 seconds.

SPECT image reconstruction

Prior to the reconstruction process, the raw data of the SPECT phantom images were corrected for Technetium-99m (Tc-99m) decay, uniformity and centre of rotation (COR). The filtered back-projection (FBP) method was used for the reconstruction of images. The Butterworth filter was used with a cut off frequency of 0.35 cycle/cm and order 5 to reduce the noise from reconstructed images. The Chang's attenuation correction was used with the linear attenuation coefficient, 0.11/cm.

Data analysis

The reconstructed images in this study were analysed quantitatively. For uniform region standard deviation in the count density was obtained using ImageJ software (Rasband 2020). A large region of interest (ROI) over the entire reconstructed image of uniform region was drawn to achieve the standard deviation. Hot and cold regions were analysed measuring the contrast and SNR. The average count density of hot/cold region (D_{region}) and background ($D_{background}$) was recorded by drawing region of interest (ROI) inside the hot/cold region carefully so that it did not cover a background region. Whereas ROI for the background region was drawn in such a way that it did not overlap the hot/cold region.

For the calculation of contrast of hot & cold regions and SNR, equations 1, 2 and 3 were used, respectively.

$$C_{hot} = \frac{D_{region} - D_{background}}{D_{region} + D_{background}} \quad 1$$

$$C_{cold} = \frac{D_{region} - D_{background}}{D_{background}} \quad 2$$

$$SNR_{hot/cold} = C \;\; \times \sqrt{D_{background}} \quad 3$$

RESULTS AND DISCUSSION

The results of and discussion on three regions, i.e., uniform, hot and cold regions have been presented as follows:

Uniform region

Uniformity of SPECT imaging systems is routinely checked to make sure the response of the system to uniformly distributed radiopharmaceutical in the object to be imaged is linear. The impact of non-uniformity on clinical images is the appearance of hot, cold areas and ring artefacts in the reconstructed image. Standard deviation is one of parameter among the others to check the non-uniformity in SPECT images. In this part of the study we investigated the influence of the phantom-to-collimator distance on uniformity. Results show about 49% increase in standard deviation with the increase in phantom-to-collimator distance which was altered from 15.9 to 18.9 cm (Figure 2). This clearly indicated that if the patient-to-gamma camera head/collimator distance is larger the adverse impact in the form of hot, cold and ring artefacts could appear on the reconstructed image which lead to the degradation of overall image quality. Consequently, this compromises the accuracy in the diagnosis of diseases.

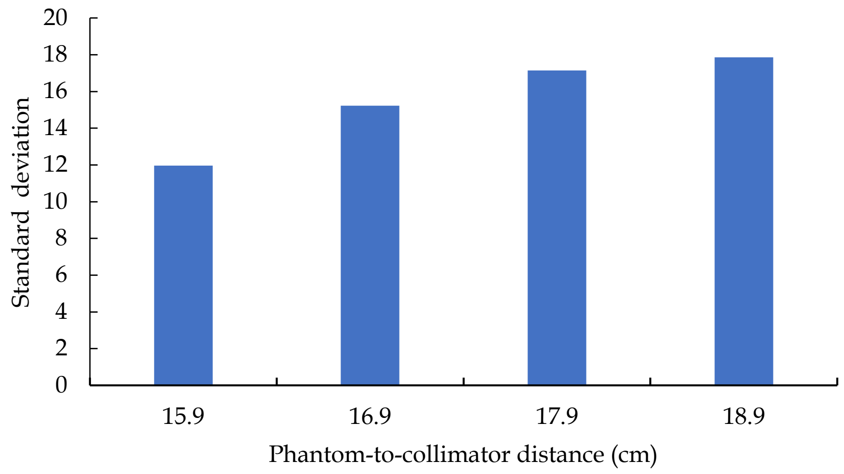

Figure 2 Standard deviation in the count density of reconstructed images of the uniform region of phantom.

Hot regions

Figure 3 represents the contrast values of hot regions at different phantom-to-collimator distances. Only three sized hot regions i.e., 22.3, 17.9 and 14.3 mm diameter were considered because of their visibility in the image and the rest of hot regions were excluded from analysis. Generally, it was found that there was a reduction in the contrast of hot regions with respect to the increase in phantom-to-collimator distance. Also, for the smaller sized hot regions the decrease in contrast was recorded. The overall decrease in contrast resulting from the increase in the phantom-to-collimator distance was noted to be approximately 1- 3%, which is considered to be small (Figure 4). On the contrary, at 17.9 cm and 18.9 cm phantom-to-collimator distance the contrast increased slightly up to 2.5% as shown in Figure 4.

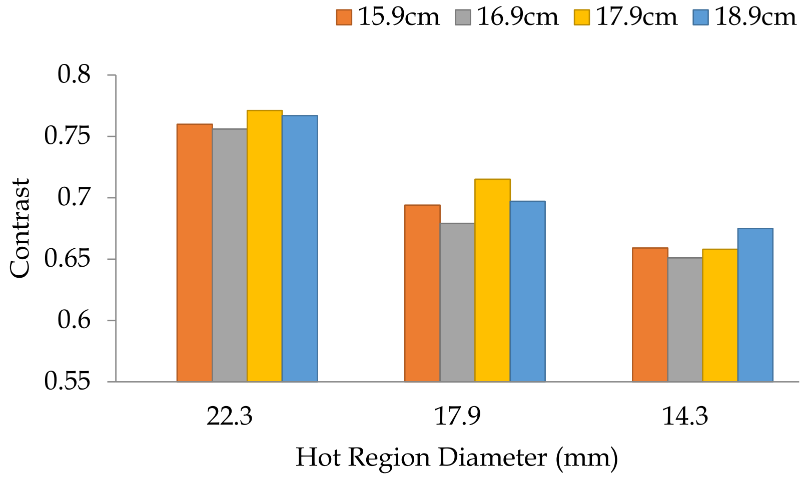

Figure 3 Contrast of various sized hot regions at different phantom-to-collimator distances.

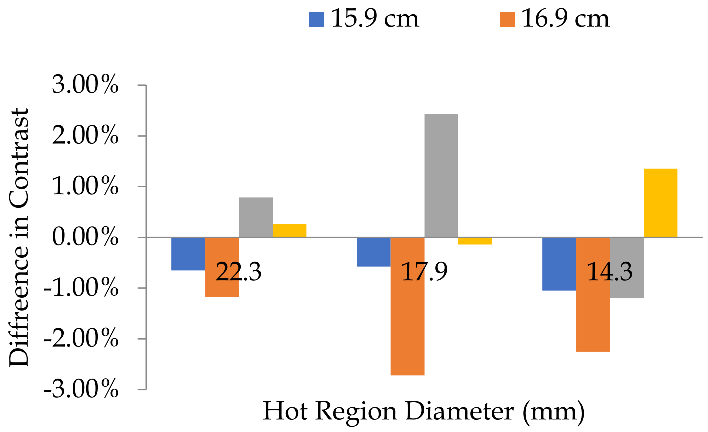

Figure 4 Comparison of the percentage difference in the contrast for hot regions at different phantom-to-collimator distances.

The difference in terms of increase in the contrast was found in fewer observations compared to the decrease in the contrast with respect to the increase in phantom-to-collimator distance. This might have resulted due to the error in drawing the region of interest (ROI) since it was a bit difficult to exactly identify the boundaries of hot regions. Thus, the inaccurate average count value measurements were attained which affected the calculated contrast values. Furthermore, the decrease in the contrast of hot regions is in agreement with what Kohli et al (1998) stated that the lesion contrast is only higher when the system resolution is improved. This is because, typically the resolution of collimator degrades by a factor of 2 at a 4 - 5 cm phantom-to-collimator distance (Cherry et al 2012).

Figure 5 illustrates the SNR values of hot regions at various phantom-to-collimator distance. It is found that the SNR of all hot regions is reduced at all phantom-to-collimator distance ranging from 5% to 31%, as shown in Figure 6.

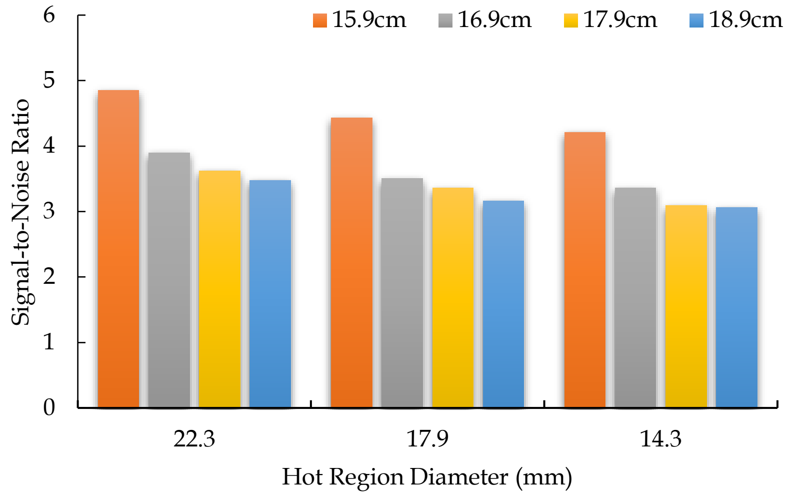

Figure 5 Signal-to-noise ratio of hot regions with respect to different phantom-to-collimator distance.

This result supports the findings of uniform region where increase in the standard deviation was found with the increase in phantom-to-collimator distance. It means that noise component increases as the phantom-to-collimator distance increases. In clinical images, when noise level is higher than the signals the grainier images are obtained and it becomes difficult in distinguishing the region of interest from its background. Ultimately, this would compromise the image quality and lead to the reduction in diagnostic accuracy (Inayatullah, 2020).

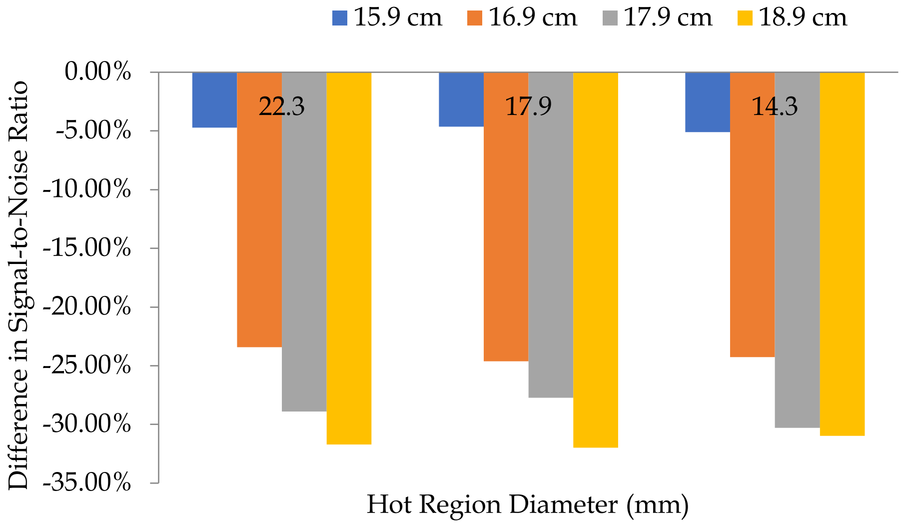

Figure 6 Comparison of the percentage difference in the SNR of hot regions with respect to various phantom-to-collimator distances.

Cold regions

Figure 7 shows the contrast values of cold regions with respect to different phantom-to-collimator distance. In this case six cold regions were detectable and included for analysis. The contrast of all cold regions decreased except 9.2mm diameter where a slight increase in the contrast at all phantom-to-collimator distance was recorded.

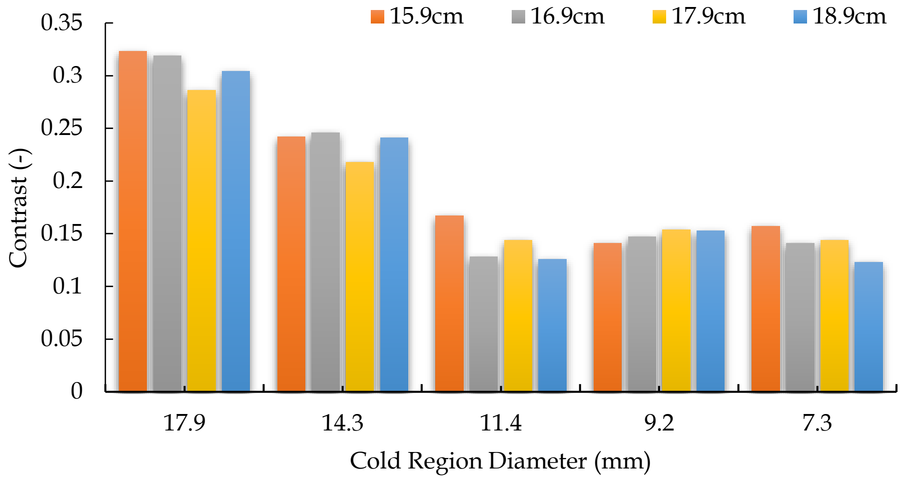

Figure 7 Contrast of different sized cold regions at various phantom-to-collimator distance

The percentage difference (decrease) in the contrast of cold regions with respect to the increase in phantom-to-collimator distance calculated ranged from 2 - 38%. However, the increase in the contrast with respect to the increase in phantom-to-collimator distance was recorded from 4 – 25% as shown in

Figure 8. The effect of low contrast images in patient studies hinder the visibility of regions under study which then compromises the accuracy in the diagnosis of diseases.

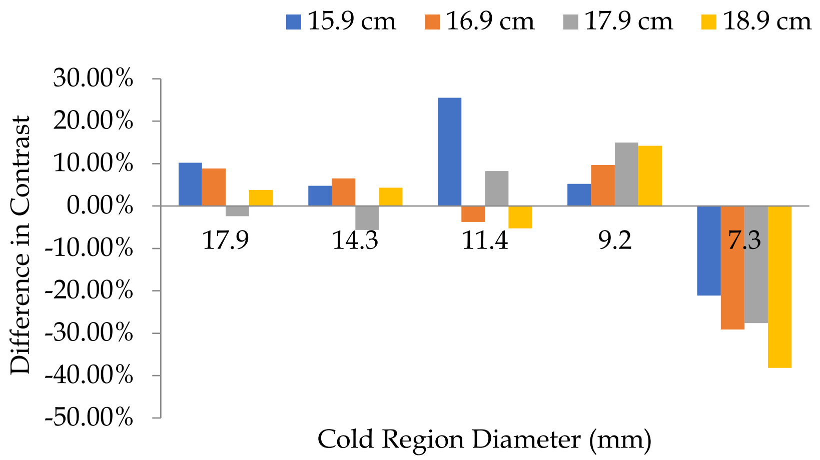

Figure 8 Comparison of the percentage difference in the contrast of cold regions with respect to phantom-to-collimator distance.

Analysis of the image quality of cold regions was done by calculating the signal-to-noise SNR at various phantom-to-collimator distances. Figure 9 represents the SNR values of cold regions. It was found that the SNR reduced for all of the cold regions when the phantom-to-collimator distance increased. The percentage difference (decrease) in SNR recorded ranged from 15 - 56% when the phantom-to-collimator distance was increased as shown in Figure 10. However, at 15.9 cm phantom-to-collimator distance an increase of 2 – 21% in SNR was observed except for 7.3 mm cold region where a decrease in SNR was noticed.

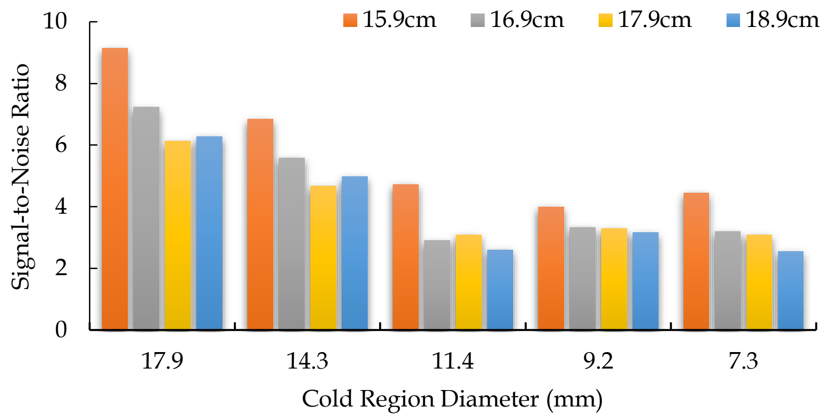

Figure 9 Signal-to-noise ratio of cold regions with respect to different phantom-to-collimator distance

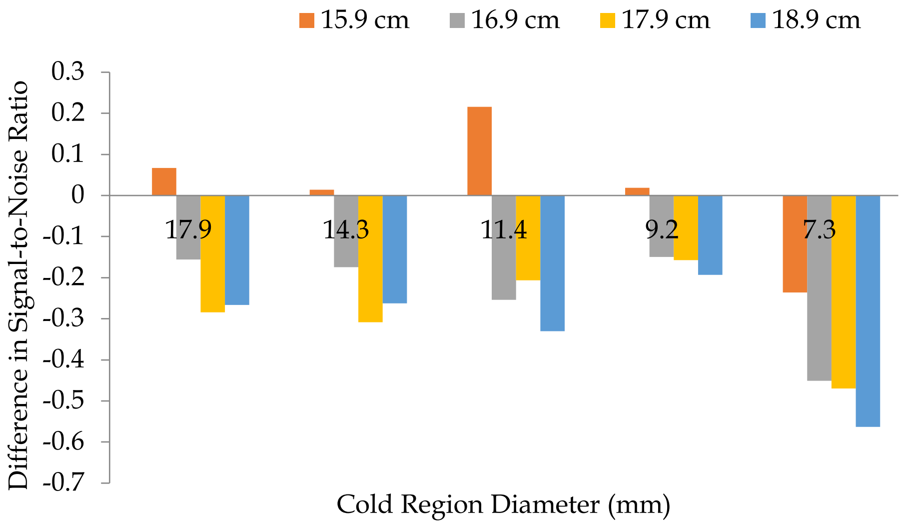

Figure 10 Comparison of the percentage difference in SNR of cold regions with respect to phantom-to-collimator distance.

CONCLUSION

In conclusion, the standard deviation in the count density of the reconstructed images from the uniform region of the phantom increased with the increase in phantom-to-collimator distance. For hot regions contrast unequivocal results were recorded. However, for cold regions a decrease in the contrast for most of the phantom-to-collimator distance was found. The signal-to-noise ratio of hot and cold regions reduced substantially with respect to the increase in phantom-to-collimator distance. Generally, the overall results indicate that the farther the object to be imaged from the gamma camera collimator, the poorer would be the quality of the reconstructed image. Therefore, in SPECT clinical examinations attention on the source-to-collimator distance needs to be given in order to achieve the diagnostic quality images.

REFERENCES

Bailey, D. L. Humm, J. L. Todd-Pokropek, A. van Aswegen, A. (Eds. (2014). Review of Nuclear Medicine Physics: A Handbook for Teachers and Student, IAEA publication (ISBN 978–92–0–143810–2).

Cherry, S.R. Sorenson, J.A. and Phelps, M.E. (2012). Physics in Nuclear Medicine (4th edn.). Philadelphia: Saunders, USA.

Frank P. DiFilippo. (2019). Assessment of PET and SPECT phantom image quality through automated binary classification of cold rod arrays. Med. Phys. 46 (8), 3451.

Glick, S.J. Penney, B.C. King, M.A. and Byrne, C.L. (1994). Noniterative Compensation for the Distance-Dependent Detector Response and Photon Attenuation in SPECT Imaging. IEEE Transactions on Medical Imaging, Vol. 13, No. 2, 363-374.

Inayatullah S. Sayed, and Siti S. Ismail. (2020). Comparison of Low-Pass Filters for SPECT Imaging. International Journal of Biomedical Imaging, vol. 2020, Article ID 9239753, 9 pages, 2020. https://doi.org/10.1155/2020/9239753

International Atomic Energy Agency. Nuclear Medicine (2013). Retrieved February 26, 2013. https://rpop.iaea.org/RPOP/RPoP/Content/InformationFor/HealthProfessionals/3_NuclearMedicine/index.htm

Kohli, V. King, M.A. Pan, T-S. And Glick, S.J. (1998). Compensation for Distance- Dependent Resolution in Cardiac-Perfusion SPECT: Impact on Uniformity of Wall Counts and Wall Thickness. IEEE Transactions on Nuclear Science, Vol. 45, No. 3, 1104-1110.

Li, J. Liang, Z. Ye, J. and Han, G. (1998). An Investigation on Analytical Methods for Correction of Distance-Dependent Resolution Variation in 3D SPECT Imaging. Paper presented in Nuclear Science Symposium, Toronto, Ont, Vol. 3, 1620- 1624

Milena, D. Sonya, S. and Aleksandra, J. (2018) Non-uniformity images - effect of NaI(Tl) Crystal Hydration. Journal of Nuclear Medicine, 59 (supplement 1) 1852.

Motomura, N. Nambu, K. Kojima, A. Tomiguchi, S. and Ogawa, K. (2006). Development of a Collimator Blurring Compensation Methods Using Fine Angular Sampling Projection Data in SPECT. Annals of Nuclear Medicine, Vol. 20, No. 4, 337-340.

Noori-Asl M. (2020). Investigation of different factors affecting the quality of SPECT images: A simulation study. J Med Phys; 45:44-51.

Rasband, W. S. (2020). *ImageJ*, U. S. National Institutes of Health, Bethesda, Maryland, USA, https://imagej.nih.gov/ij.