The variation of intrinsic spatial resolution across the UFOV of scintillation cameras

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Abstract

The aim of the present study was to investigate in detail the variation of the intrinsic spatial resolution across the useful field of view (UFOV) of \(/^9_{\text{Tc}}H\) cameras and to explore whether this variation could lead to observable effects in clinical images. Two \(/^9_{\text{Tc}}H\) cameras were used, without their collimators, to acquire 560 \(^{99m}\)Tc point source images at different points across their UFOVs, in order to measure the intrinsic spatial resolution at each point. Possible clinical effects of the resolution variation were examined on images of a thyroid phantom using a LEHR collimator, acquired at different locations on the UFOV and at various distances from the collimator. The \(^{99m}\)Tc-camera intrinsic resolution varied significantly across the UFOV, being generally lower at the central region and deteriorating at the edges. Pronounced local maxima and minima were found at points corresponding to the centers of the photomultiplier tubes (PMTs) and halfway in between. Maximum differences of more than 50% were observed between the points presenting the best and worst intrinsic resolution. Differences between neighboring points reached 17%. The effects of resolution variation were clearly observable on the thyroid phantom images.

It was concluded that an appropriate correction algorithm might be necessary in order to correct for the variation of the intrinsic spatial resolution across the UFOV of \(/^9_{\text{Tc}}H\) cameras.

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1. Introduction

The ongoing development of diagnostic and, especially, therapeutic applications in nuclear medicine requires acquisition of images containing accurate quantitative information about the radiopharmaceutical distribution inside the patient’s body. Unfortunately, a series of factors that cause serious degradations in the quality of \(/^9_{\text{Tc}}H\)-camera images inhibit the achievement of this target. The most important of those factors include photon attenuation, photon scatter and distance-dependent spatial resolution [1–5]. Significant efforts have been devoted over the previous years in the development of efficient correction algorithms in order to restore image quality, some of which have already been incorporated into clinical practice [6–11]. There is, however, one parameter that degrades the quality of the images obtained with the \(/^9_{\text{Tc}}H\)-cameras, which has not been extensively studied in the literature: the variation of \(/^9_{\text{Tc}}H\)-camera spatial resolution across the useful field of view (UFOV). The aim of the present study was to investigate in detail the variation of the intrinsic spatial resolution across the UFOV of \(/^9_{\text{Tc}}H\)-cameras, providing an initial step towards the development of an appropriate correction method.

2. Materials and methods

Measurements were performed on two single-head SPECT gamma cameras. The first was an ADAC Genesys camera, installed in 1994 at the Nuclear Medicine Department of the AHEPA University Hospital, Thessaloniki, Greece, with a 50.8 cm × 38.1 cm rectangular field of view, equipped with 55
round photomultiplier tubes (PMTs) (49 with a 3″ diameter and 6 with a 2″ diameter) arranged on a hexagonal array. The second was a digital head Sopha Medical Vision (SMV) DS7 camera, installed in 2001 at the Interbalkan Medical Center, Thessaloniki, Greece, with a 40 cm diameter circular field of view, equipped with 61 hexagonal PMTs (size 6 cm) arranged on a hexagonal array.

Intrinsic resolution measurements were performed using a collimated point source phantom (Fig. 1), which was constructed by drilling a 1 mm diameter hole through the base of a cylindrical lead container. The lead phantom had a height of 70 mm and an outer diameter of 55 mm. The thickness of the lead walls was 14 mm, providing an attenuation factor of the order of $10^{-17}$ to the 140 keV $^{99m}$Tc photons (considering a HVL value of 0.25 mm). A glass vial filled with $^{99m}$Tc eluate was placed inside the container. The activity of the radioactive solution into the glass vial was 370–740 MBq (10–20 mCi) in a volume of 2–4 ml. The $^{99m}$Tc eluate was taken from a freshly eluted 12 GBq CIS Biointernational 99-Mo/$^{99m}$Tc generator.

Images of the point source were acquired at 430 points across the UFOV of the SMV DS7 gamma camera. Fig. 2 shows a representation of the measurement positions across the two gamma cameras UFOVs. The circles mark the positions of the PMTs. The acquisition protocol used in all measurements is shown in Table 1. The intrinsic energy resolution (FWHM) for the 140 keV $^{99m}$Tc photons was 10.6% for the ADAC camera and 9.8% for the SMV DS7 camera. All acquisitions were performed with all available correction matrices enabled (linearity correction, energy correction, spectrum correction and flood correction). The linearity correction matrix had been acquired less than 4 months before the onset of the acquisition of the presented data whereas the energy, spectrum and flood correction matrices had been acquired less than one month before.

The point source images were processed further in order to generate three-dimensional Point Spread Functions (PSFs). A rectangular $21 \times 21$ pixels region of interest (ROI) was centered on each point source image, covering the entire count distribution around it. An ASCII table was generated for each ROI which contained the counts acquired in each pixel inside it. Each table was then fed into a PC based three-dimensional regression software (TableCurve 3D v2, Jandel Scientific Software) where the measured pixel counts were fitted with a Gaussian function $G(x, y)$ of the type:

$$G(x, y) = a \exp \left( \frac{-(x - b)^2}{2e^2} \right) \exp \left( \frac{-(y - d)^2}{2e^2} \right)$$

where $a$ is the distribution peak counts, $b$ the $x$-axis centroid ($x_0$) in pixels, $c$ the $x$-axis sigma ($\sigma_x$) in pixels, $d$ the $y$-axis centroid ($y_0$) in pixels and $e$ is the $y$-axis sigma ($\sigma_y$) in pixels. $G(x, y)$ represents the intrinsic PSF($x, y$), from which the intrinsic spatial resolution can be determined.
resolution $R_i$ at each point $(x_0, y_0)$ can be determined from the full width at half maximum (FWHM) along the $x$- and $y$-axis from the relations:

$$R_{i,x} \text{(mm)} = \text{FWHM}_x = 2\sigma_x \sqrt{2\ln 2} w$$
$$R_{i,y} \text{(mm)} = \text{FWHM}_y = 2\sigma_y \sqrt{2\ln 2} w$$

(2)

where $w$ is the appropriate pixel size factor, in order to express spatial resolution in mm. For the ADAC Genesys and an acquisition matrix of $512 \times 512$, $w = 1.21$ mm/pixel (the SMV DS7 provided all measurements directly in mm).

The possible effects of resolution variation across the UFOV on clinical images were investigated by imaging a standard thyroid phantom (Picker 4205) filled with 3.7 MBq (100 $\mu$Ci) $^{99m}$Tc solution. Four nodules were incorporated into the thyroid phantom: a 6 mm diameter cold nodule on the upper right lobe, a 12 mm diameter cold nodule on the lower right lobe, a 9 mm diameter cold nodule on the upper left lobe and a 12 mm diameter hot nodule on the lower left lobe. Images were acquired on the ADAC Genesys with a LEHR collimator, using the acquisition protocol used in everyday clinical practice (acquisition time: 300 s; matrix size: $256 \times 256$; zoom factor: 2.19) at three phantom-to-collimator distances, namely 0, 7 and 10 cm. Two images were acquired at each distance: at the first, the phantom position on the UFOV was such that the 6 mm cold nodule on the upper part of its right lobe was aligned with the center of the central PMT, whereas at the second the same cold nodule was placed half-way between the central and its two neighboring PMTs (Fig. 3).

3. Results

Fig. 4 shows a representative example of an intrinsic PSF at a point on the UFOV of the ADAC Genesys obtained after fitting the measured counts inside the ROI of the point source with the Gaussian function $G(x, y)$.

<table>
<thead>
<tr>
<th>Phantom-to-collimator distance</th>
<th>Lobe at PMT center</th>
<th>Lobe between PMTs</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 cm (%)</td>
<td>40.7</td>
<td>43.6</td>
</tr>
<tr>
<td>7 cm (%)</td>
<td>24.6</td>
<td>31.2</td>
</tr>
<tr>
<td>10 cm (%)</td>
<td>19.5</td>
<td>25.0</td>
</tr>
</tbody>
</table>

Table 2 Values of cold lobe image contrast (%) for each image of the thyroid phantom

Figs. 5–9 present results from the ADAC Genesys. They show the variation of the intrinsic resolution $R_i$, in terms of its components $R_{i,x}$ and $R_{i,y}$, along a central vertical line (Fig. 5), along a central horizontal line (Fig. 6) and along two diagonal lines of the UFOV (Figs. 7 and 8). Fig. 9 shows values of $R_i$ at points directly upon the center of the PMTs and half way in between.

Figs. 10 and 11 present the measured values of the intrinsic resolution $R_i$, along a central vertical line (Fig. 10), along a diagonal line (Fig. 11) of the SMV DS7 UFOV.

Fig. 12 presents the three pairs of images of the thyroid phantom acquired at phantom-to-collimator distances of: (a) 0 cm, (b) 7 cm and (c) 10 cm. On the upper image of each pair the cold nodule at the upper right lobe was aligned with the center of a PMT, whereas at the lower image the cold nodule was placed halfway between three neighboring PMTs. Fig. 13 shows vertical line profiles along the cold nodule in each image. Circular ROIs were placed at the center of the cold nodule and the surrounding lobe in order to measure image contrast in each image, using the relation:

$$C = \frac{|Y_{lob} - Y_{nod}|}{Y_{lob}} \times 100$$

(3)

where $Y_{lob}$ and $Y_{nod}$ are the corresponding counts inside the lobe and the nodule ROIs. Table 2 shows the calculated image contrast values for each image.
4. Discussion

The intrinsic resolution of both cameras has been found to vary significantly across their UFOV. Intrinsic resolution was generally better at the central region of the UFOV and deteriorated towards the edges. This finding is sometimes observed in quadrant bar phantom images acquired during quality control measurements: the thinner bars which can be resolved...
at the central region of the UFOV cannot be resolved at the edges.

In ADAC Genesys, the lowest measured intrinsic resolution values were $R_{i,x} = 3.7$ mm and $R_{i,y} = 3.6$ mm and the highest were $R_{i,x} = 5.7$ mm and $R_{i,y} = 5.7$ mm. This represents a 54% variation in $R_{i,x}$ and a 58% variation in $R_{i,y}$. Significant variations have also been observed between neighboring points inside the UFOV: $R_{i,x}$ varied by 16% (from 3.7 to 4.3 mm) and $R_{i,y}$ by 17% (from 3.6 to 4.2 mm). It should be noted that the lowest $R_{i,x}$ and $R_{i,y}$ values were not measured at the same point. Generally, intrinsic...
resolutions at x and y axis ($R_{i,x}$ and $R_{i,y}$) were not strongly related to one another (correlation coefficient 0.409).

The improvement of the intrinsic resolution from the edges of the UFOV towards its center was not monotonous; both $R_{i,x}$ and $R_{i,y}$ showed pronounced local maxima and minima. All the local maxima appeared at points corresponding to the centers of the PMTs and all the minima at points halfway in between. Therefore the intrinsic resolution deteriorated at points corresponding to the centers of the PMTs, improved at points halfway in between and varied smoothly from one local extreme to

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**Fig. 8.** Variation of the intrinsic spatial resolution ($R_{i,x}$ and $R_{i,y}$) of the ADAC Genesys along an other diagonal line of its UFOV.

**Fig. 9.** Measurements of the intrinsic spatial resolution ($R_{i,x}$ and $R_{i,y}$) of the ADAC Genesys directly upon the centers of the PMTs and halfway in between.
the next. To our knowledge, this finding has not been previously studied extensively in the literature. A simple description, without any further analysis, was only found in a recent IAEA publication [12].

A similar (although less pronounced) behavior of the intrinsic resolution was also found in SMV DS7: a general improvement was observed towards the center of the UFOV which was not monotonous but, instead, presented local maxima and minima.
corresponding to the centers of the PMTs and to points halfway in between. The overall variation was found to be 13% (both $R_{i,x}$ and $R_{i,y}$ varied from 3.1 to 3.5 mm) whereas local variations between neighboring points reached 10% (from 3.1 to 3.4 mm).

The poorer $\gamma$-camera resolution at the edges of the UFOV is attributed to the lower light collection efficiency of scintillation events that occur at the specific regions of the crystal, which is due to the smaller number of PMTs surrounding these events. On the other hand, the significant variation in resolution between neighboring points located centrally on the crystal does not seem to be due to light collection variations. It seems more likely that such variations could be due to differences in the relative errors of the PMT signals used by the positioning circuitry. A more detailed explanation is attempted below.

The intrinsic spatial resolution of a gamma camera is mainly determined by two factors: the Poisson fluctuation of the number of photoelectrons in each PMT contributing with its signal to the positioning circuitry and the PMT response function (PRF) which is defined as the PMT counts versus point source position. When a scintillation event takes place at a position aligned to the center of a PMT, the signal of that specific PMT is the highest, but the slope of the PRF for the PMT at that position is flat, making that PMT pretty useless to determine the exact
scintillation position. It is the neighboring PMTs—with steep PRF slopes at that position but with much lower signals (with larger Poisson fluctuations)—that play a dominant role in position analysis. On the other hand, when the scintillation occurs between 2 or 3 PMTs they will all contribute significantly to the position determination, offering 2 or 3 relatively large signals (with less Poisson fluctuations) and possibly even steeper PRF slopes than the previous case, leading to a more accurate determination of the exact scintillation position and, therefore, to a better intrinsic spatial resolution at that region.

The magnitude of the resolution variation seems to be significant enough to result in observable degradations in the quality of clinical images. The images of the thyroid phantom suggest that the location of a feature inside the imaged object with respect to the $\gamma$-camera PMTs can affect its imaging quality and, in special cases, its detectability. The image contrast of the cold nodule improves, as its location with respect to the PMTs changes from directly at the PMT center to half-way between two PMTs. At the distance of 10 cm, the lobe is no longer detectable when it is located at the PMT center but it is clearly visualized when located between PMTs.

It can be concluded that the variation of the intrinsic spatial resolution across the UFOV of modern $\gamma$-cameras could be a significant source of degradation in both planar and tomographic images. Appropriate correction algorithms might be necessary in order to account for this effect.

5. Summary

The spatial resolution of scintillation cameras is determined by the intrinsic resolution of the scintillation crystal and its electronics and the collimator resolution, which is known to be strongly dependent on the source-to-collimator distance. The intrinsic resolution is presumed to be constant across the UFOV, with the exception of its outer edges, despite the fact that there are no detailed measurements to support this. The results of 560 Point Spread Function measurements at two SPECT $\gamma$-cameras show that intrinsic spatial resolution varies significantly across the UFOV, being worse exactly below the centers of the PMTs and improving in between. Appropriate correction algorithms may be necessary to account for this effect, which could affect the quality of clinical images.

References


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