LUNG COUNT RATE ESTIMATIONS USING A PORTABLE SCINTILLATION DETECTOR PRIOR TO V/P_{SPECT}

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Abstract

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Purpose: The aim of this study was to find a method using a portable scintillation detector to predict the tomographic image statistics after inhaling technegas prior to V/P SPECT, and to validate the method for clinical use.

Background: V/P SPECT is a tomographic imaging (SPECT) method using $^{99m}$Tc labeled carbon particles (technegas) and $^{99m}$Tc-MAA, for examination of the lung ventilation and perfusion, respectively using a gamma camera. Technegas is inhaled and a sufficient count rate is important in the ventilation images. A too low count rate gives inadequate image quality and a too high count rate can lead to difficulties when constructing quotient images between ventilation and perfusion images. To verify an acceptable count rate the local clinical routine is to use the camera detector. This method increases e.g. the risk of contaminating the camera head because of its air cooling system. It would be advantageous preparing the patient in another room using a portable detector; saving camera time, lower the risk of contaminate the camera and to see the count rate in real time.

Method: A portable scintillation detector (SSL Radhound with SS404 Al probe Southern Scientific UK, via Gamma Data AB) was used and its characteristics like rotational symmetry, laterally size of detection field, distance dependence and the best placement of the detector was investigated. The contribution from the technegas in patient administration set (PAS) is relatively high, and must be considered since only the count rate from the administrated technegas in the patient is of interest. Therefore an additional lead shield is placed on the detector. The positioning of the detector and the optimal height from the patient to the detector was investigated. The measurements were made with and without a PAS as well as with and without additional lead shielding to obtain a restricted field-of-view (FOV). To verify this method the number of counts in the diagnostic ventilation images was correlated to the portable detector count rate.
Result: From measurements, the detector crystal was symmetric in response. The lateral size of the shielded FOV was a circle with a radius of 20 cm. From these investigations the method of choice was to use an additional lead shield, and place the detector 20 cm above the xiphoid process with the patient lying on the examination bed in supine position. The correlation between the diagnostic ventilation images and the portable detector count rate implies that a count rate of approximately 14 kcps is favorable to achieve adequate statistic in the tomographic images, when the patient administration set was removed. To obtain the recommended tomographic image statistics within a 95% confidence interval, the total number of counts was calculated to be 800-2500 kCts, and then the count rate from the portable detector should be 12-15 kcps.

Conclusion: Based on all tests performed in this project the most promising clinical method for this portable detector was determined. The portable detector should be placed on a fixed height 20 cm over the patient’s xiphoid process. The count rate directly after the inhalation should be approximately 15 kcps. When the patient administration set is removed and the count rate is controlled again a sufficient value of the count rate is then 13-15 kcps. If the count rate is too low additional inhalation is needed.
Foreword

This Master of Science Thesis of 30 hp was written for my degree in Medical Physics at the University of Gothenburg. The work was performed during the spring semester of 2015 at the Nuclear Medicine Department at Sahlgrenska University Hospital.

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Abbreviations

$^{99}$Mo  Molybdenum 99
$^{99m}$Tc  Technetium 99 metastable
$^{99m}$TcO$_4^{-}$  Pertechnetate
°  Degree
%  Percent
ADC  Analog to digital converter
°C  Degree Celsius
cm  centimetre
cps  counts per second
e.g.  Exempli Gratia (for example)
eV  Electron volt
FOV  Field-of-view
GM-tube  Geiger Müller tube
LEGP  Low Energy General Purpose
h  Hour
HAS  Human Serum Albumin
i.e.  Id est (that is)
keV  kilo electron volt
kcps  kilo counts per second
kCts  Kilo counts
LEHR  Low Energy High Resolution
µm  micrometre
MAA  Macro Aggregated Albumin
MCA  Multi-Channel Analyzer
MBq  Mega Becquerel
MeV  Mega electron volt
ml  millilitre
mm  millimeter
NaI(Tl)  Sodium Iodine Thallium activated
OSEM  Ordered Subset Expectation Maximization
PAS  Patient Administration Set
PE  Pulmonary Embolism
PM-tubes  Photo Multiplication tubes
SU  Sahlgrenska University Hospital
V/P$_{\text{SPECT}}$  Ventilation/Perfusion Single-Photon Emission Computed Tomography
VTE  Venous Thromboembolism
Z-value  Atom number
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Introduction

Ventilation/perfusion single-photon emission computed tomography (V/P$_{SPECT}$) is a tomographic nuclear imaging method using gamma cameras for diagnosing pulmonary diseases, mainly pulmonary embolism (PE). PE is when a blood clot remains in the lung. The most common symptoms for PE are chest pain and dyspnea. A difference between the genders can be seen where hemoptysis, chest pain and unilateral edema of the lower leg has been reported for men while women reports calf pain, increased anxiety and dyspnea [1]. The incidence of PE is approximately 100 cases per 100,000 people per year and is equally distributed between the genders. The risk of PE for young people, especially children under 15 years old is low but it increases dramatically after the age of 60. To be able to detect PE a visualization of the lung function consisting of both ventilation and perfusion is needed. The first image encounters the ventilation meaning the examination of the gas exchange between the lung tissue and the atmosphere. The image is made by inhalation of a radioactive gas for example technegas, and the recommended number of counts in the image is 2000 kCts [2]. The second image shows the perfusion which is the blood stream through the vessels in the lung made by injection of $^{99m}$Tc-labeled macro aggregated albumin (MAA). To be able to see if there are aberrations of the lung function the perfusion imaging takes place directly after the ventilation imaging with the patient remaining in the same position. This makes it possible to make quotient images after subtracting the ventilation images from the perfusion image and then compare it to the ventilation image. In parts of the lung where there is low perfusion but sufficient ventilation PE is suspected. Since the isotope of use is the same in both the ventilation and perfusion image the count rate for the ventilation must be sufficiently high but not too high otherwise there can be trouble detecting aberrations in the quotient image [2].

At Sahlgrenska University hospital (SU) the gamma camera detectors are used for determination if the inhaled count rate from the technegas for ventilation examination is sufficient. Since the technegas is airborne there is a great risk of contamination on the camera detector especially when inhaling near the camera head because of its air-cooling system, which can easy receive the technegas to the detectors and cause contamination. The patient is therefore placed on the bed as far away from the gamma camera-detectors as possible during the inhalation and then moved into the gamma camera to validate the count rate. In some cases the count rate is too low and the patient must move to the initial position to inhale more activity. This is a problem because the technegas must be inhaled within ten minutes after creation since the particles come together which leads to clusters. The increased size could otherwise lead to that the particles stop before reaching the alveolus where the gas exchange occurs [2]. It is therefore important to establish a sufficient count rate as fast as possible. Patients with PE might have trouble to inhale the technegas which increase the risk of getting a too low count rate to have a good image quality. On contrary, patients with no trouble inhaling the gas can have a too high count rate.

There would be advantages to prepare the patient in another room using a portable detector; this would both save camera time, lower the risk of contamination of the camera room, and make it possible to see the count rate in real time. The purpose of this study was to solve these problems by developing a clinical routine to use a portable scintillation detector for count rate measurements of the inhaled technegas and relate it to the total number of collected counts in the gamma camera detectors. There are no articles published using this type of detector for this application and therefore it needs to be documented. Therefore the main focus of this study was the ventilation part of the V/P$_{SPECT}$ examination. During V/P$_{SPECT}$ imaging there are three main components affecting the examination: the patient, the gamma camera and the radioisotopes. A more detailed description of these components and the aim of this study is presented in the following text.
Respiratory system

The lungs are placed in the chest cavity surrounded by the thorax. They are divided into lobes, which make the lungs more mobile during the ventilation. The left lung consists of two lobes and is smaller than the right lung which consists of three lobes, Figure 1.

The cells in the human body have a mechanism called metabolism which is necessary to generate energy to be able to perform vital functions. In the metabolism oxygen is required and must be transported into the cells in the body tissues. This is made by the blood trough the vascular system, which can be divided into two parts, the systemic circulation system and the pulmonary circulation system [3]. The systemic circulation transports oxygenated blood to the body tissue and brings back carbon dioxide rich blood to the right camber of the heart. In this chamber the pulmonary circulation starts and during this circulation the gas exchange takes place.

![Figure 1](image.png)

A schematic image in posterior view describing the composition of the lungs. The right lung consists of three lobes: the superior lobe, the middle lobe and the inferior lobe while the left lung consists of two lobes: the superior lobe and the inferior lobe.

After inhalation, air reaches the lungs via the trachea which branches out to the bronchus and further to bronchioles to finally reach the alveolus. From the right heart camber the blood is transported by the pulmonary arteries to smaller and smaller capillaries until it reaches the alveolus, where the gas exchange occurs. The carbon dioxide changes to oxygen in the blood which is then transported back to the left heart chamber and then further out to the cells in the body. The carbon dioxide follows the airways out from the body by the expiration [4].

If the lungs do not function correctly as for example when having a blood clot, an emboli, in one of the larger vessels in the lung the oxygenation of tissue in other parts of the body can be insufficient which can lead to necrosis. The emboli can occur in several ways. It can occur from a broken vessel fixed by coagulated blood, if a fraction of this separates from the wound and does not resolved as it should. Another way is formation by reduction of the normal blood circulation. A third is when the blood composition changes [5].

Venous thromboembolism (VTE), which is a formation of blood clots in the veins, developed in the lower extremist is a major risk factor in developing PE [1]. This can be seen as a signal loss in the perfusion images, since the blood clot stops the blood flow in the pulmonary capillaries but it will still be a signal in these areas for the ventilation image when a V/F SPECT is performed [2].
Risk factors for VTE can be divided into three parts according to Virchow’s triad. The first part is venous stasis and involves arterial fibrillation, immobilization, long-distance travel and paralysis [1]. The second part is endothelial injury as hypertension, indwelling catheter, surgery and trauma. The third and last step is hypercoagulability and contains malignancy, obesity, sepsis and smoking. Other factors affecting VTE are acute medical illness and advanced age. In some cases, mostly seen in younger women, PE is induced by contraceptives. The risk increases when using combined oral contraceptives [6].

The emboli increase the pulmonary vascular resistance which can lead to hypertension and the workload on the right ventricle increases, right-sided heart failure can therefore accrue. Due to hypertension and pulmonary obstruction the sympathetic nervous system releases chemical mediators that cause vasoconstriction. This causes a decrease in blood flow to the pulmonary vasculature and that gives a mismatch in V/P SPECT images. The PE patient might still have good ventilation but not good oxygenation due to the obstruction. A greater mismatch leads to a higher decrease in production and this could lead to atelectasis and pulmonary edema. Ischemia and vasoconstriction can cause an irreversible damage to the lung and heart and can even lead to a heart infarction. Due to this it is important that the disease is treated as soon as possible [1].

**Gamma camera and scintillation detectors**

In general a gamma camera consist of a collimator, a sodium iodide thallium activated (NaI(Tl)) scintillation crystal, light guide, photo multiplication tubes (PM-tubes) and analog to digital converters (ADC). Figure 2 is a schematic figure of the organization of these components.

![Gamma Camera Components](image)

**Figure 2**

A schematic image over the gamma camera components. The gamma camera consists of a collimator, a NaI(Tl) crystal, a light guide, PM-tubes and ADCs.
To generate a good image quality the scattered photons approaching the camera must be stopped. This is made by the collimator which consists of lead, so called septa, which absorbs these photons. Depending on the organ or tissue that is aimed to be visualized the septa are angled differently, there are four types of collimators: parallel hole, pinhole, converging and diverting collimator. A parallel hole collimator is the most clinically used collimator in nuclear medicine and it is used for V/P$_{SPECT}$ imaging. The lead is placed in a hexagonal pattern, Figure 3, and the thickness of the septal wall is based on the energy of the photons of use. For example low energy high resolution collimator (LEHR) is mostly used for photons with energies smaller than 150 keV and it has a thin septal wall compared to collimators used for higher photon energies. The image created with this collimator will be in natural size [7].

![Parallel Hole Collimator](image3.png)

Figure 3
The hexagonal pattern of the lead walls in a parallel hole collimator.

NaI(Tl) is the most common crystal used in inorganic scintillators and has a good linearity and light output. During gamma examinations inorganic scintillators is to prefer because of their high Z value i.e. high particle density leading to good absorption properties. Other advantages are that it has a low self-absorption, is transparent for its own scintillation emission and can be made in large sizes relatively cheap.

The main purpose of the crystal is to convert energy from photons into visible light [7]. This is made by the incoming photons which transfer their energy to electrons in the scintillator crystal. In a pure crystal the de-excitation of the electrons becomes ineffective since the broad bandwidth gives photons that lie outside the visible spectra. Therefore small impurities of thallium, called activators are added to the scintillators. The activators give energy stages creating more visible photons. The generation of light from NaI(Tl) is approximately 38 000 photons per MeV energy deposition [8].

The light guide transports the scintillation photons from the NaI(Tl) crystal to the PM-tube by its optical couplings with the two components [7].

In a gamma camera there are several PM-tubes placed in a hexagonal pattern to cover as large area as possible of the crystal [7]. The purpose of a photo multiplication tube is to convert the light signal to a current to be able to read out the number of pulses collected. A typical light pulse from the scintillator only contains a few hundred photons which is a too small number to be a sufficient pulse and must be enlarged. The electron multiplication part of the PM-tube is by its geometry a good amplifier for photoelectrons and increases the number gently to $10^7$-$10^{10}$ electrons. To create an excited electron in the dynode material the energy must be at least the size of the bandgap. Electrons in the dynode are exited from energy rich electrons passing through the material and give the multiplication portion of the PM-tube. It is theoretically possible to create 30 exited electrons per 100 eV accelerating voltage. The output stage or anode collects the charge [8]. The scintillation detector used in this report consist of a NaI(Tl) scintillation crystal and a PM-tube previously described. The scintillation crystal is shielded internally with a 3.15 mm thick lead collimator [9].
To be able to discriminate scattered photons we are only interested in the full energy peak. This can be made by using pulse height spectrometry, which is used for investigation of the signal amplitude to only incorporate the signals in a predetermined interval. The detector output must be proportional to the detected radiation energy as for the scintillation detector. A pulse height spectrum displayed as the number of detected events versus the amplitude of these events, is needed to be able to analyze the pulse height. The easiest way of making such spectrum is to use a Multi-Channel Analyzer (MCA). The MCA is an analog to digital converter with several channels divided into different discrete intervals of pulse amplitude. The MCA then summarizes the numbers of pulses with certain amplitude and creates a spectrum, Figure 4.

Figure 4
An example on how a spectrum for $^{99m}\text{Tc}$ can appear. The full energy peak is located at 140 keV.

At high count rates there can be aberrations in the spectrum due to the overlap between detector output pulses for example between photo peak events and low energy events this creates a broadening of the peak. One explanation of this pulse pile up is the dead time in the detector. The dead time is the time between the detection of the photon in the detector and the pulse produced in the detector. There are two types of dead time: paralyzing and non-paralyzing systems. If there is an event during the dead time in a non-paralyzing system, this event will be ignored. An event during the dead time will extend the dead time for a paralyzing system, even if the event is not counted. A scintillator is a paralyzing system and most of the dead time is located in the pulse amplifier and in the analyzer. The true count rate is higher than the measured because of the dead time [7].

V/P\text{SPECT} is a tomographic method meaning that the gamma camera detectors rotate and collects image information, called projections, from the photons allocating from the radioactive isotope, at various angles around the body. The interval of collection angles and the time for collecting the projections are predetermined. A reconstruction algorithm then transforms the projections into images. The algorithm used at SU for V/P\text{SPECT} is Ordered Subset Expectation Maximization (OSEM). This algorithm uses sets of projections called subsets and iterates them with Expectation Maximization using projection followed by its inverse back projection. OSEM are thanks to its subsets a method where the convergence is accelerated with a factor proportional to the number of subsets [10]. An iteration process is when the recorded projections are compared to a “guessed” projection that often is blank or uniform from start. In the next iteration step the forward projections are used to calculate the projections that would have been measured for the estimated image. In the forward projection all the potential radiation rays trough the estimated image are summarized and afterwards they are compared
with the actually recorded profiles. Then by using the difference between the guess and the recorded projection the guess are updated and a new comparison between the projections takes place and the iteration process converges towards the true image. The number of iterations is ether predetermined or depends on a specific value for which the difference should lie below. The drawback with iterative reconstruction algorithms it that it is relatively time consuming [7].

Radiopharmaceutical

At SU patients with suspected PE can be diagnosed by V/P_{SPECT} in which technegas is used for the ventilation imaging and $^{99m}$Tc-MAA is used for the perfusion imaging. Both radiopharmaceuticals consist of the metastable isotope $^{99m}$Tc which is a radionuclide daughter from $^{99}$Mo isotope emitting gamma radiation, Figure 5. $^{99m}$Tc has a half-life of 6.02 h and an energy peak of 140 keV [11]. The mother isotope, $^{99}$Mo is created in nuclear reactors as a product of the fission of $^{235}$U and $^{99m}$Tc is produced by a generator. In the generator $^{99}$Mo has the form of an ion $^{99}$MoO$_4^{2-}$. Therefore it can bind to an aluminum column. $^{99m}$Tc is then produced in the form of $^{99m}$TcO$_4^{-}$ (pertechnetate) which does not have such strong binding to the aluminum and can be eluted from the column by using a saline solution. The activity of $^{99m}$Tc is rebuilt and reaches its maximal amount 24 h after elution, but the $^{99m}$Tc can be used three to six hours after an elution. A commercially prepared generator is used one week and is then discarded because of the natural decay of the $^{99}$Mo [7].

![Figure 5](image)

**Figure 5**

A simplified decay scheme of $^{99}$Mo to $^{99m}$Tc and $^{99}$Tc. The energy of the mean beta particles are 0.4426 MeV.

Technegas

Technegas is aerosols which consist of two parts ($^{99m}$Tc and carbon) and it is airborne and relatively stable over time. The particle size is 0.005-0.2 µm but can increase due to aggregation (formation of clusters) and therefore the technegas should be used within ten minutes after creation [2].

Technegas is produced in an aperture, which works like a small high temperature furnace. It uses a combination of graphite and argon atmosphere to vaporize the pertechnetate first by reducing the water in the pertechnetate and then heating the rest. Afterwards the temperature in the aperture rises to 2750 °C for 3-15 seconds and the technegas particles are produced [12].

The number of technegas particles in the lung tissue depends on the particle features, mainly the particle size. For very small particles, up to 50 % of them remain in the lungs. Due to the diffusion mechanism, i.e. the motion of particles from a high concentration to a lower concentration of particles,
most technegas particles are deposed in the alveoli. The number of particles deposed in the lungs decrease to half when the particle diameter is larger than 0.1 µm. When the diameter has increased to 1 µm, most technegas particles are deposed in the lower airways. If the particle size is larger than 5 µm the major part of the technegas particles stops at the upper respiratory tract. Another factor that is important for the ventilation is the respiratory pattern. Relatively large particles can reach the lung periphery at slow tidal breathing [2]. This technique is therefore used in diagnostics for the ventilation function, where for example radiolabeled aerosols can be used.

99mTc-MAA
99mTc-MAA has a particle size of 10-45 µm and can be made by neutralization of heat treated human serum albumin (HAS). The HAS denatures fully and then 99mTc-microspherer can bind to the HAS. [13]. The number of particles injected are about 400 000 but since the lungs have 280 billion pulmonary capillaries the blocking of these vessels will not affect the function of the lung. If a patient has pulmonary hypertension (i.e. high pulmonary pressure) the number of particles have to be considered and recommended number of particles are 100,000-200,000 to avoid blocking of pulmonary capillaries [2].

Aim
The aim of this study was to find a method using a portable scintillation detector to predict the tomographic image statistics after inhaling technegas prior to V/P SPECT, and to validate the method for clinical use.
Material and methods

The portable detector used for all measurements in this report (SSL Radhound with SS404 Al probe, Southern Scientific UK, via Gamma Data AB) is referred to as “detector” in the following text. Two gamma cameras were used for the V/P SPECT imaging, a SPECT/CT with a diagnostic CT (Discovery NM/CT 670, GE Healthcare, Sweden), and a SPECT/CT with low dose CT (Infinia Hawkeye NM/CT hybrid GE Healthcare, Sweden) the detectors of these cameras are referred as “gamma camera detector” in the following text.

Establishment of the clinical method using a portable detector

Detector characteristics

To see if the count rate has a lateral dependence and if the response is symmetric, the count rate from a reference point source using the detector was evaluated by the following concept. The detector was placed in a clamp attached to a stand standing on a table. The height from the table to the probe was 9 cm. A point source of 0.1 ml $^{99m}$Tc with an activity of 20 MBq was placed directly under the detector and then moved from the detector along the table, in steps of 5 cm at a time at first along the z-axis and the x-axis, Figure 6. A correction for the natural decay was made for all measurements using:

$$r = r_0 e^{\frac{\ln(2) \times t}{T_{1/2}}}$$

(1)

in equation 1 $r_0$ is the count rate measured when preparing the point source at time $t=0$, $r$ is the count rate, $T_{1/2}$ is the half-life of $^{99m}$Tc (6.02 h) and $t$ is the time passed since the measurements started.

The same procedure was repeated with the detector at a height of 20 cm form the table, but only along one of axis. An additional lead shielding was used as extra collimation and the measurements were repeated for both heights. The lead shielding had a thickness of 2 mm and hung down 1 cm around the probe, Figure 7. A correction for natural decay for the count rate was made by using equation 1. These values were normalized by dividing each series of count rates with count rate when the point source was centered under the detector respectively.

![Figure 6](image_url)

Schematic figure of the placement of the detector and the direction of choice for the x- and z-axis.
Dead time effects was investigated by placing the detector probe, with lead collimation described in Figure 7, on a stand 20 cm above the table. A $^{99m}$Tc source with a start-activity of 39.5 MBq in a volume of 0.3 ml was placed on the table centered under the detector. A computer was placed on the table so that its webcam could be used to take pictures of the detector display. A webcam program (WebCamImageSave, Nir Sofer, http://www.nirsoft.net) was used, allowing the images to be captured every third second during the measurement. Every whole hour during 15 hours an average value of the count rate was calculated from the four first count rates observed by the webcam image. The activity of the point source was corrected for natural decay using:

$$A = A_0 e^{-\frac{\ln(2) \cdot t}{T_{1/2}}}$$

In equation 2, $A_0$ is the start-activity when preparing the point source at time $t=0$ (39.5 MBq), $t$ is the time passed since the first measurement, $A$ is the activity at time $t$ and $T_{1/2}$ is the half-life of $^{99m}$Tc (6.02 h).

To verify the dead-time impact on measured count rate, calculations was performed applying the following equations [8]:

$$r_c = \frac{r_m}{(1 - r_m \tau)}$$

where $r_c$ is the corrected count rate, $r_m$ is the measured count rate and $\tau$ is the dead time (10 $\mu$s) [9], and use it in the following equation since it is a paralyzing system:

$$r = r_c e^{-r_c \tau}$$

Where $r$ is the true count rate, $r_c$ is the corrected count rate calculated by equation 3 and $\tau$ is the dead time (10 $\mu$s) [9].
Clinical measurements
To find a method suitable for clinical use different detector settings was investigated. The first clinical measurements were performed with the detector placed directly at the patient’s chest. Five different locations were measured in the following order: the upper left lung, the lower left lung, the xiphoid process, the upper right lung and the lower right lung, visualized in Figure 8.

![Figure 8](image)

**Figure 8**
Placement in the clinical measurement according to the patient’s lungs Index 1 is the upper placement of the left lung, 2 is the lower placement of the lung, 3 is the xiphoid process, 4 is the upper right lung and 5 is the lower placement of the right lung.

The activity contribution from the patient administration set (PAS) was determined by placing it on the patient bed in the same way as the patient would have it, Figure 9, with 24 cm from the end of the mouthpiece to the center of the detector. The detector with the lead shield, previously described, was placed in the same position corresponding to the xiphoid process as it would be during an examination. The count rate was measured at a detector height of 42 cm above the table. The detector was then moved closer to the table with measuring points two centimeters apart. Then the same series of measurements were repeated without the lead shield. The count rate was corrected for the decay time from the inhalation to the measurement according to equation 1.

![Figure 9](image)

**Figure 9**
The patient during the ventilation with the administration set and the technegas aperture.
According to the results from previous measurements in this project a clinical method was developed. To verify the method it was applied and evaluated on 21 patients with suspected PE. The portable detector was placed at the same distance (20 cm) above the xiphoid process during the clinical preparation for the ventilation SPECT. The scintillation detector was shielded as described in Figure 7. The count rate was measured with and without the PAS present, and was compared to the total number of collected counts during the ventilation SPECT. The gamma camera detector count rate was also compared to the total number of counts during the ventilation image. Statistically validation of the method was made.

**Statistical analysis**

An average value of the count rate was calculated by using:

\[ \bar{x} = \frac{\sum x_i}{n} \quad (5) \]

Where \( x_i \) is the count rate of the portable detector for measurement \( i \) and \( n \) is the total number of measurements.

A standard deviation was calculated for the total number of counts by:

\[ SD = \sqrt{\frac{\sum (y_i - y)^2}{n - 1}} \quad (6) \]

Where \( y_i \) is the total number of counts in the ventilation image from patient \( i \), and \( y \) is the total number of counts calculated by using a linear trend line fitted to the data points collected from all patients in the study (21 patients), \( n \) is the total number of measurements.

A 95% confidence interval was calculated from using the equation:

\[ \bar{x} \pm 1.96 \times SD \quad (7) \]

A linear trend line was fitted to the data points using the method of least squares to have the best fitted trend line with the smallest distance to the data points. The \( R^2 \) value says, if the \( x \) value is known, with which certainty \( y \) can be predicted. The value of \( R^2 \) vary from 0.0 to 1.0. If it is close to 1.0 the trend line is a good approximation and from a certain \( x \) value we can predict the corresponding \( y \) value with certainty. If the value on the other hand is close to 0.0 the prediction of the \( y \) value is vague for a known \( x \) value.
Results

Establishment of the clinical method using a portable scintillation detector

Detector characteristics
The detector response was similar in all direction according to the point source measurements described in Figure 6, the results, Figure 10, is displayed as a function of the distance from the centrum of the detector.

Figure 10
The count rate dependence of the lateral dislocation is displayed in steps of five centimeters from the center of the detector. The dark blue dots represent the count rate along the z-axis and the light blue dots display it along the x-axis. The activity used for the point source was 20 MBq and the height from the detector to the table was 9 cm.

The portable detector response at a point source-to-detector distance of 9 cm or 20 cm was normalized and resulted in a steeper slope for measurements 9 cm above the point source than 20 cm above, Figure 11. Therefore the FOV is smaller for the shorter distance and has a radius of 20 cm, but the count rate was 60 % lower of the full count rate when measuring at a lateral dislocation larger than 10 cm, and 10 % lower of the full count rate with a distance larger than 15 cm. The slope was also steeper for the shielded detector when the point-source to detector distance was 20 cm. From the measured profile the radius of FOV for the shielded detector was estimated to be 35 cm. The count rate decreases until 60 % with a lateral dislocation larger than 15 cm and to 9 % with a distance of 30 cm. The radius for the unshielded detector at a source-detector distance of 20 cm the FOV was larger than 60 cm. Less than 60 % of the full count rate was measured at a lateral dislocation of 15 cm and less than 10 % is measured at a distance of 50 cm. The normalized count rate for both the measurements of a height of 20 cm was the same until a lateral distance of 20 cm (corresponding to the size of the lung) then the count rate for the shielded detector decreases faster than the unshielded. These results indicates that it was preferable to have a source-detector distance of 20 cm instead of 9 cm to avoid signal loss from the lungs, and collimation should be used to obtain an adequate discrimination of the signal from the PAS.
The normalized count rate dependence of the lateral dislocation in steps of five centimeters from the detector. The blue dots represent count rate measured with a lead shielded detector, detector-source distance of 9 cm. The red and the green dots are measurements without and with shielding, respectively, with a detector-source distance of 20 cm.

Count rate increases with activity, Figure 12. The $R^2$-value shows good correlation of the measuring points with the linear trend line. Using equation 3, and equation 4 the calculation indicates that the dead time, according to manufactures specifications was maximal 23 % lower for the measured values for the clinical relevant activity that is in the range of 20-25 MBq.

The measured count rate is displayed as a function of activity (◊). A linear trend line was fitted to the measured points and the equation for this line plus the $R^2$-value is displayed in the upper right corner of the figure. A count rate corrected for the dead time is displayed (□).
Clinical measurements

PAS had large impact on the count rate when measuring with a portable detector without shielding. This indicates that a lead shielding is necessary if the detector is situated closer to the patient than 34 cm where similar measurements were received for the shielded and unshielded detector, Figure 13. Up until a distance of 20 cm the count rate for the shielded detector was low and forms a plateau, after that the count rate to the shielded detector increased rapidly until a distance of 36 cm. At distances wider than 34 cm the count rate for the unshielded detector was lower than for the shielded.

As expected, the count rate measurements using gamma camera detectors was well correlated to the total number of counts in the tomographic ventilation images, Figure 14a, while the correlation between the tomographic ventilation image total counts and the portable detector count rate was weaker. A count rate of 13.6 kcps using the portable detector gave a total number of 800-2500 kCts (95 % confidence interval) during the ventilation imaging.
Figure 14

(a) Count rate during inhalation measured using a gamma camera detector positioned below the patient compared to the total number of counts during the tomographic ventilation imaging.

(b) Total number of collected counts during the ventilation imaging displayed as a function of the portable detector count rate without the PAS present in the field. The blue line represents the average count rate of 13.6 kcps and the interval between the purple lines represent the 95% confidence interval for this count rate.

The data from the clinical measure 20 cm above the xiphoid process without the PAS in the field was fitted to a linear trend line with the equation:

\[ y = 121.3 \times x \quad (8) \]

where \( y \) represents the total number of counts during the ventilation images and \( x \) is the corresponding count rate.

There are larger fluctuations between measured count rate and the total number of counts when the PAS is present in the measuring field. A calculation of the deviation in percent from the contribution of the PAS was made by using:

\[ \text{contribution} = \left( 1 - \frac{r}{r_{\text{PAS}}} \right) \times 100 \quad (9) \]

in this equation \( r \) is the count rate without the PAS and \( r_{\text{PAS}} \) is the count rate with the PAS present in the FOV.

When measuring a higher count rates the contribution from the tube were larger. An average of the deviation close to previously calculated value of 13.6 was calculated. This was made by summation the count rates in the interval of 12-15 kcps and divide it with the number of count rates (five) used. This gave a value of 1.0 kcps.
Figure 15

(a) Represent the total number of collected counts during the ventilation imaging is displayed as a function of the portable detector count rate with the PAS present in the FOV. A linear trend line was fitted to the measuring points and its equation plus the $R^2$ value can be seen in the lower right corner.

(b) Represent the amount of scattered radiation in percent whit the measured count rate from the portable detector when the PAS is present in the FOV.

Suggestion of clinical method

From all measurements the most suitable clinical method for this detector was determined. The detector should be placed on a fixed height of 20 cm over the patient’s xiphoid process. The PAS should be placed as far away from the detector as possible, preferably at least 15 cm from the detector. The count rate directly after the inhalation should be approximately 14 kcps using the portable detector which corresponds to a statistics in the tomographic image around 2000 kCts. Afterwards the PAS is removed and the count rate is controlled again. A sufficient value of the count rate is 13-15 kcps. If the count rate is to low additional inhalation is needed.
Discussion

There are several different techniques to measure the count rate from the technegas in the lungs. There are two detector types suitable for ventilation measurements; the scintillation detector and the Geiger Müller tube (GM-tube). In a study by Palmer et al a GM-tube was used to determine the activity in the lungs at the ventilation [14]. In another study by K. Tägil et al a GM-tube was used to determine if the activity to lung was sufficient at the ventilation. The principle was to place the 25 cm long, shielded detector over the patient in supine position. There was a slit in the shielding lead at the placement of the third rib. When reaching a count rate of 1000 cps, corresponding to an activity of 20 MBq, the inhalation was interrupted [15]. The GM-tube is well documented for these applications while the use of scintillation detector has to our knowledge not previously been published.

The use of GM-tubes in nuclear medicine is limited due to the stopping power, the low efficiency of x-ray and gamma radiation, and the dead time for GM-tubes is longer (typically 50-100 µs) than for most other detector systems [8]. An inorganic scintillation detector on the other hand has better efficiency [7]. The scintillator also has similar components as a gamma camera detector since it also consist of a NaI(Tl) crystal and a PM-tube which can make it easier to relate them to each other. In e-mail conversation with another clinic in Sweden it appeared that they had used a GM-tube but the detector was unstable and slow in response. The result was that the patient in some cases inhaled too much activity before the detector stabilized. When considering these difficulties a scintillation detector seemed like a good choice.

The type of scintillation detector investigated in this report was already in clinical use before this project started. Two clinics in Sweden were asked about their method of using the scintillation detector. Both methods involved measuring in the transversal plane on the “side” of the patient. One of the clinics always measured at the right side of the patient and the other clinic measured at either the right or the left patient side, just below the armpit. The advantage of measuring from the side is that the contribution from the activity in the PAS decreases due to the internal shielding of the detector crystal in the probe. The distance to the lungs is also smaller when measuring on the side. The disadvantages are that the contribution from the lung closest to the detector will have a larger influence on the detector which could be a problem if the function for this lung is low compared to the count rate from the other lung for example when having only one lung. The FOV of the detector is also smaller when measuring closer to the patient. This could lead troubles when crating images since the amount of inhaled activity is more or less than what it is expected to be. The method investigated in this report was performed with the detector placed in front of the patient in the coronal plane as well as using a lead shielding around the detector to avoid contribution from the PAS. It would be favourable to measure with the same distance to the activity (i.e. the lungs) for each patient but in practise it is hard since the distance to the lungs from the chest wall varies with each patient, therefore a fixed distance from the patient to the detector is used.

The examination of the lateral dislocation dependence of the detector was performed to see if there was any aberration in the detector response in different directions from the detector. The count rate decreases with the lateral dislocation up till 40 cm. With a larger distance than this the count rate is small and therefore the FOV is assumed to be a circle with a radius of 20 cm for an unshielded detector with a source-detector distance of 9 cm. When comparing the count rate in Figure 10 for the same distance, some aberrations can be seen. Especially for the lateral dislocation of 5 cm the different is rather large. The differences in count rate for the measuring points can be due to the fluctuation of the displayed detector value. Another possible source of error is that the count rate was rounded into three significant numbers. It can also depend on if the detector was placed completely horizontal without rotation around any of the axis, if not this will give some difference s in distance from the detector to the source and also an asymmetric FOV. This would result in an increased amount of count rate in the measurement points close to the detector. The placement of the source can vary in the different directions since the source was encapsulated by a shoot and is therefore not completely
spherical in shape. The variations were relatively small and when considering the possible sources of errors, the detector response is said to be symmetric and independent of the rotation.

When looking at Figure 11 for the source-detector distance of 9 cm the amount of the total normalized count rate decreases much faster than for the same measuring point with a detector-source distance of 20 cm. Since the count rate is more equally distributed and the FOV is larger for a source-detector distance of 20 cm it seems more suitable to place the detector at this distance. The count rate for both the measurements of a height of 20 cm is the same until a lateral dislocation of 20 cm (corresponding to the size of the lung) then the count rate for the shielded detector decreases faster than the unshielded. This indicates that the lead shield is a good collimation for the detector to decrease the FOV but still have a sufficient size. These results indicates that it is preferable to have a source-detector distance of 20 cm, to avoid signal loss from the lungs, and using collimation to obtain an adequate discrimination of the signal from the PAS.

A possible source of error is the dead time since the detector will become paralyzed. It is therefore a risk of inhaling too much technegas which results in a too high count rate before the count rate stabilizes. An investigation to simulate the dead time affects in the patient situation for the shielded detector with a source-detector distance of 20 cm was performed. A small source with a count rate a bit lower than the highest measured count rate in the patient situation was measured. The $R^2$-value of the lineal trend line was close to one, but a trend where the sensitivity decreases with the activity can also be seen. If the dead time was affecting the count rate there would be a plateau at high count rates in Figure 12. Differences between the corrected count rate and the measured count rate is relatively small but increases rapidly with the activity. By using equation 3 and equation 4 for a dead time of 10 $\mu$s the dead time plateau, theoretically, starts at around 100 kcps. The specifications of the detector underestimates the dead times for the detector. In the clinical case the dead time is not especially important as long as the detector is not paralyzed since we are interested to find a value corresponding to a sufficient activity with or without dead time present. The correlation between count rate measured with the portable detector and the tomographic statistics in the final image includes effects of dead time, and no further corrections are needed.

When placing the detector probe close to the patient in the first pilot measurements, there were large fluctuations between the different locations and the various patients. This variation could be due to the patient’s ability to ventilate and the difference in the patient’s anatomy. For example a taller patient often has longer lungs than a shorter. The activity distribution will therefore be in a smaller FOV for the shorter patient. The placement of the probe at the patient can also contribute to the differences in count rate, due to the decrease in the FOV when placing the probe closer to the patient as previously discussed. The count rate was lowest for the xiphoid process because of the increased distance to the lungs compared to measuring directly above one of the lungs. When increasing the distance from the patient to the detector it seemed suitable to place the detector so that the contribution in count rate from both the lungs is as equal as possible. The detector was therefore placed above the xiphoid process even though it is in the lower part of the lungs. This is advantageous because the distance to the PAS is lager and the sternum is not directly under the probe which could otherwise contribute to decrease in count rate due to its rather high density and attenuation, compared to soft tissue. Another advantage is that it is relatively easy to locate the xiphoid process and it is therefore a good mark for the placement of the detector.

Since the measurements are preferred in real time to get a hint of the inhaled count rate from the technegas, the PAS will be present and contribute to the count rate. In an attempt to minimize this contribution count rate measurements were made on a used PAS placed in the same way as a patient would have it. The detector was placed at different heights from the examination table, with and without lead shielding to account for the different field sizes and detector responses at various heights and shielding. When looking at Figure 13, count rates for the lead shielded probe were lower than the count rates from the unshielded probe at a distance smaller than 32 cm. For the unshielded probe the
count rate increased with the height to 24 cm from the PAS but the increase was slower compared to
the lead shielded detector. The shielded probe had a plateau between the values of 10-15 cm and then
the count rate raised rapidly with the distance until 36 cm. This indicates that a suitable detector to
patient distance is within 10-15 cm even though a detector-source distance until 32 cm seem suitable if
using lead shielding. The distance from the patient to the detector used during patient measurements
was 20 cm even though this distance according to the measurements of the contribution from PAS
indicates that the height should be less. If a detector setting in the interval of 10-15 cm was used, as
the PSA measurements indicated there would be a risk that the tips of the lungs would not be within
the FOV for tall patients.

From the patient measurements with a lead shielded detector placed 20 cm above the xiphoid process,
an average count rate of 14 kcps for the portable detector was observed to be a suitable count rate. To
assure that the patient will have a sufficient image quality for the average value a 95 % confidence
interval was mad. This interval encountered the total number of counts and was 800-2500 kCts. A total
number of 2000 kCts gives according to Palmer et al a suitable image statistics during a tomographic
image collection if inhaling $^{99m}$Tc-DTPA, and a ventilation image set for 128 projections, a10 s in 64
steps using OSEM as reconstruction algorithm with eight subsets and two iterations. At SU the same
iteration principle is used but the collection time is a bit longer and the collimator is a LEHR
compared to a general purpose (LEGP) collimator used by Palmer et al [14]. The result of the shorter
collection time and the use of another collimator will be a lower number of the total collected counts,
approximately half the number, which gives more noise in the images. A number of collected counts
for a LEHR is expected to be half of the number form the GP collimator during the same collection
time. The value corresponding to 2 000 kCts for a LEGP collimator is therefore 1 000 kCts for a
LEHR collimator. The lowest count rate in the 95 % confidence interval is 800 kCts and this is a bit
low but rather close to the desired minimum value of 1000 kCts and therefore accepted. When the
patient is in the higher part of the interval close to 2500 kCts the statistics will be better but it can be
hard making good quotient images. To obtain the recommended tomographic image statistics the
portable detector count rate of 12-15 kcps was observed. If the number of counts in the ventilation are
too high a solution is to adjust the amount of the intravenous injected $^{99m}$Tc-MAA. The count rate for
the $^{99m}$Tc-MAA should be four times the count rate of the technegas and can be achieved by reading
out the count rate of the gamma camera detector for the ventilation at first, and then the perfusion by
slow injection. In the clinic there is almost impossible to have exact the average value and there are
also uncertainties when measuring the count rate due to for example detector fluctuations therefore an
interval from 13-15 kcps is accepted as a sufficient count rate since it gives some trading margin.
There are large fluctuations in count rate for different patients, this can depend on when (in time) the
count rate was measured. It can also depend on the patient size which leads to different distance from
the lungs to the detector and a different attenuation of the radiation, these parameters effects the
portable detector more than the statistics in the tomographic image.

The contribution from the PAS seemed to depend on the count rate, Figure 15. A higher count rate
gave a larger PAS contribution. To be able to get a hint of when it is time to stop inhaling the patient
an estimate of the contribution for the count rates, which was near the average count rate of 14 kcps,
measured when the PAS was present in the FOV. The contribution was around 10 %. When
compensating for this the count rate was 15 kcps and can be used as an estimation value for when the
count rate is in the range of the requested count rate. There are tough hard to see if the activity is
sufficient in real time, since the contribution from the PAS is still too high and varies from patient to
patient. To be sure of the count rate from the technegas in the lung PAS must be taken away or this
contribution must be lower. Further investigations of other methods is therefore needed. The count rate
must therefore be adjusted to the accurate count rate of 13-15 kcps when the PAS is not present in the
FOV.

When comparing the method previously used, Appendix 1, with the method suggested in this report
the $R^2$ value in Figure 14 shows better correlation for the previously used method then for the
suggested. This could be to the fact that the detector used for both the determination of sufficient count rate and the ventilation imagine is the same and therefore the correlation seems better. Another possibility is that the time from the inhalation to the measurement with the camera detector is longer than for the measurement with the portable detector and therefore the number of administrated particles is more stable than directly after the inhalation. Another factor is that the FOV is larger of the gamma camera detector than for the portable detector. This leads to that the estimation of larger lungs for taller patient will be better for the camera detector. The count rate for the portable scintillation detector is approximately ten times as large compared to the count rate for the gamma camera detector. This is due to the collimation and the window settings in the gamma camera which limits the number of detected counts and therefore a lower count rate can be expected for these measurements. This also contribute to a more certain value of the gamma camera detector measurements. Despite the better $R^2$ value for the previously used method there are other advantages of the suggested method. Some of the advantages are that the patient can inhale technegas in another room which saves camera time and lower the contamination risk. All the preparation with the technegas aperture can also be made in the same place. It is more convenient for the patient for which the majority is old and has difficulties breathing when lying flat on their back since the time in the positon decreases since they do not need to go back and forth under the camera detectors. It is also convenient for the staff since it is possible to get a hint of how well the patient can inhale the technegas in real time and therefore the risk of a too high or low count rate decreases. Since the aim of this study was not to improve the method, rather it was to find a method to a portable scintillation detector to predict the tomographic image statistics after inhaling technegas prior to $V/P_{SPECT}$. The method suggested in this report, Appendix 2, fulfills this aim and can replace the previously used method.
Conclusion

Based on all tests performed in this project the most promising clinical method for this portable detector was determined. The shielded portable detector should be placed on a fixed height 20 cm over the patient’s xiphoid process. The count rate directly after the inhalation should be approximately 15 kcps. Afterwards the PAS is removed and the count rate is controlled again, a sufficient value is 13-15 kcps. If the count rate is to low additional inhalation is needed. In the clinical validation the suggested method fulfills the following aspects. The technegas can be inhaled in a room without a camera and all the preparation with the technegas aperture can be made in that room. This saves camera time and lowers the contamination risk of the camera. It is more convenient for the patients for which the majority is old and has difficulties breathing when lying flat on their back since the time in the position during inhalation decreases. It is convenient for the staff since it is possible to get a hint of how well the patient can inhale the technegas in real time and therefore the risk of a too high or low count rate decreases.
**Future aspects**

This method needs further development to be optimal for clinical application. The next step would be to find a suitable time from the inhalation to the measurement and to use the suggested method and to validate it by using the “old method” as a control of the count rate.

Some problems still need a solution in the further development of the method and they are the count rate variation with time, the contribution of counts from the PAS, and the variation of patient parameters that affects the lung function. The most affecting patient factors are according to Ostrowski et al. the height and gender but the age also has an impact of the lung function [16]. These issues should be examined further to find the optimal method for clinical use.

A more certain value for the count rate can be assured when monitoring. An average value could then be found by computer calculations or by integration of the count rate over a specific time. This would avoid stochastic fluctuations. A longer cable is needed to connect the detector probe with the monitor to place it closer to the computer. This also makes it easier for the staff to perform the examination. It even makes it possible to do it by one person instead of two persons as today since one must hold the monitor by hand.

To be able to see the count rate in real time the contribution from the PAS must be lower. The current method has decreased the count rate for some of the patient till an acceptable level for measuring the count rate in real time but for some patients the contribution is still very high. A lower count rate could be made by having a lead shield that goes further down the detector or by placing the probe closer to the patient. The drawback with this is that the FOV will be smaller. Another way of doing it is to investigate the possibility of measuring form the side of the patient previously discussed. The third option is to make a lead shield for the PAS, to cover most of the mouthpiece where most of the activity is trapped in a filter. A fourth option is to place the detector under the bed and correct for the attenuation in the bed.

The patient parameters needs to be taken into account and to do that, additional measurement are needed. A suggestion is to first measure the patients inhalation ability by using spirometry and then relate this to a standard from phantom measurements as described by Palmer et al. [14]. The patient height must also be measured. By doing this the count rate can be individually adjusted for each patient by using a correction factor. Another possibility of optimizing the examination for each patient can be made by using the gamma camera detector to estimate a count rate for the inhaled technegas. Then by slowly injecting the $^{99m}$Tc-MAA, the count rater should be increased so that the quotient between the new count rate minus the count rate measured for the ventilation, and the ventilation count rate is four. This gives the optimal image quality for making the quotient images [14].

To get better count statistics closer to 2 000 kCts recommended by Palmer et al. [14] the LEHR collimator used at SU could be replaced with a LEGP collimator.

A possibility to avoid the problem in the future is to buy gamma cameras where the cooling of the camera is not made by an air-cooling system and therefore there are no problem with contamination of the gamma camera detector when inhaling near the camera head.
Reference list

Appendix 1

V/P\textsubscript{SPECT} protocol SU

Indication and contra indication
The protocol is used for patients with suspected acute or chronical pulmonary embolism. It can also be used to diagnose obstructive pulmonary disease where other diagnostics have failed.

The amount of particles injected at perfusion is decreased if the patient: is a child, has a right-left shunt, pulmonary hypertension or a right camber pressure between 35-40 mmHg or pore lung function. If the patient is pregnant planar images with perfusion is taken as a first step. The responsible doctor then decides if the ventilation image is needed or not.

Radio pharmacy and principle
500 MBq pertechnetate and carbon particles are evaporated in the technegas aperture. The calculated dose to the lungs is approximately 20 MBq. The $^{99m}$Tc labeled carbon particles are inhaled and are mainly deposed in peripheral airways. At severe obstructive lung disease, the deposition can also accrue in more central airways. For the perfusion 140 MBq $^{99m}$Tc-MAA is administrated intravenously.

The examination
The patient is placed in supine position and inhales the technegas and then holds the breath for approximately five seconds so that the particles have time to allocate in the lung. The patient is then placed under the gamma camera detectors and the count rate should be approximately two kilo counts per second (kcps) (1.5 -2.5 kcps is accepted). If the count rate is too low the procedure above is repeated till the count rate is in the range. If possible the patient’s arms are placed above the head, if not they are crossed on the stomach. The lungs are then centered in the image field.

For the perfusion the $^{99m}$Tc are administrated slowly intravenous at the same time the patient takes three to four deep breaths and the patient remains in the same position under the detector.

The matrix size of the examination is 64 x 64 the window is 20 % the rotation is 360° (three degrees per angle) for ventilation: ten seconds per angle (total time 13 minutes) and for perfusion: five seconds per angle (total time of nine minutes).
Appendix 2

“New” V/P\text sub{spect} protocol SU

Indication and contra indication
The protocol is used for patients with suspected acute or chronical pulmonary embolism. It can also be used to diagnose obstructive pulmonary disease where other diagnostics have failed.

The amount of particles injected at perfusion is decreased if the patient: is a child, has a right-left shunt, pulmonary hypertension or a right camber pressure between 35-40 mmHg or pore lung function. If the patient is pregnant planar images with perfusion is taken as a first step. The responsible doctor then decides if the ventilation image is needed or not.

Radio pharmacy and principle
500 MBq pertechnetate and carbon particles are evaporated in the technegas aperture. The calculated dose to the lungs is approximately 20 MBq. The $^{99m}$Tc labeled carbon particles are inhaled and are mainly deposed in peripheral airways. At severe obstructive lung disease, the deposition can also accrue in more central airways. For the perfusion 140 MBq $^{99m}$Tc-MAA is administrated intravenously.

The examination
The patient is placed in supine position and the detector with additional lead shielding is placed 20 cm above the patient’s xiphoid process in the chain in the roof. The PAS is placed as far away from the detector as possible. The patient inhales the technegas and the count rate is under observation when the count rate is around 15 kcps the inhalation is interrupted. Then the patient holds the breath for approximately five seconds so that the particles have time to allocate in the lung. The PAS is then removed from the patient and placed on the technegas apparatus. The count rate of the detector is controlled and should preferably be around 14 kcps (a count rate of 12-15 kcps is accepted. If the count rate is too low the patient must inhale more technegas and afterwards a new control is made without the PAS present in the FOV. When having a sufficient count rate the patient is placed under the gamma camera detector. If possible the patient’s arms are placed above the head, if not they are crossed on the stomach. The lungs are then centered in the image field and the image collection begins.

For the perfusion the $^{99m}$Tc are administrated slowly intravenous at the same time the patient takes three to four deep breaths and the patient remains in the same position under the detector.

The matrix size of the examination is 64 x 64 the window is 20 % the rotation is 360° (three degrees per angle) for ventilation: ten seconds per angle (total time 13 minutes) and for perfusion: five seconds per angle (total time of nine minutes).