Collection and analysis of patient and image data for calibration of a voxel-phantom based Monte Carlo code and for the modelling of important structures

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Table of Contents

1. Introduction

2. Materials and Methods
   2.1. Image acquisition
   2.2. Absorbed dose measurements
   2.3. Imaging system characteristic measurements
   2.4. Image digitisation, calibration and transfer
   2.5. Image analysis
      2.5.1. Discussions with radiologists
      2.5.1.1. Chest Images
      2.5.1.2. Lumbar Spine Images
      2.5.2. Measurements in digitised images
      2.5.2.1. Contrast
      2.5.2.2. Dynamic range
      2.5.2.3. Noise

3. Results and discussion
   3.1. Patient absorbed doses
   3.2. Imaging system characteristics
   3.3. Image analysis
      3.3.1. Low-contrast detail objects
      3.3.1.1. Chest
      3.3.1.2. Lumbar spine
      3.3.2. High contrast detail objects
      3.3.3. Dynamic range
      3.3.4. Noise
   3.4. Structures selected for inclusion in the Monte Carlo model

4. Summary and conclusions
Acknowledgegements
References
1. Introduction
The contribution of the Medical Physics Departments at Linköping University (LKP) and The Royal Marsden NHS Trust (RMH) to the joint project ‘Predictivity and Optimisation in Medical Radiation Protection’ is in modelling of the chest and lumbar spine radiographic examinations. This involves:

1. the development of quantitative imaging requirements;
2. an investigation of the effect of imaging technique on image quality and patient dose, and
3. an optimisation of system design.

One of the objectives for this first reporting period (0-12 months) was to collect a set of chest and lumbar spine radiographs of patients for subsequent analysis in order to establish patient doses and important features in the images. The set of radiographs and the outcome of the image feature analysis will during this project’s second year be used to calibrate our Monte Carlo computational model of the conventional chest and lumbar spine screen-film X-ray imaging systems.

2. Materials and Methods
2.1. Image acquisition
Since it was clear that images from the first trial would not be available in time for this stage of the work, we decided to acquire our own images and arrange for them to be digitised. Fifty-eight chest images (29 patients and 2 views) were obtained in July 1996 at a neighbouring hospital (Vrinnevi Hospital in Norrköping) and in October forty-two lumbar spine images (21 patients and 2 views) were obtained at the University Hospital in Linköping. Efforts were made to ensure that equal numbers of both projections (PA and LAT view) were collected and that both male and female patients were represented.

2.2. Absorbed dose measurements
Both the entrance surface dose on the patient and the kerma-area product were measured for each patient. The kerma-area product (KAP) meter (Type No. 5753(B) PTW-Freiburg) was calibrated as described by Larsson et al. (1996) and mounted on the x-ray beam collimator.

Thermoluminescence dosimeters (TLD) of LiF (discs with 4.5 mm diameter and 0.9 mm thick) were used and read-out in TLD system DOSACUS (Alnor Oy, Finland). The TLDs were individually calibrated. The average standard variation in the calibration constant, estimated from three successive calibrations, was 1.4%. An ionisation chamber (3.7 cm³, Shonka) traceable to the BIPM was used in the absolute calibration of the dosimeters which was performed free-in air and using the same tube potential as in the subsequent use of the dosimeters in the clinic (75 kV lumbar spine and 140 kV chest).

In order to derive dosimetric quantities that are better related to radiation risk, the KAP-meter reading ($\int K_{c,air} dA$) was corrected for the fraction of the incident beam not
impinging on the patient and converted to energy imparted (ε) using conversion factors (ε/\(K_{c,\text{air}}\)dA) from Alm Carlsson et al. (1984).

2.3. Imaging system characteristic measurements
The imaging system characteristics recorded and measured were: X-ray generator type, tube potential, filtration, half value layer, HVL, focus-film distance, table-top (patient support) thickness, anti-scatter grid (ratio, lead strip width and frequency, cover and interspace material), screen-film type and sensitivity, cassette front material and thickness and film development conditions. The HVL was measured using the Solidose instrument (RTI Electronics) that measures the air-kerma. The maximum tube potential and exposure time was measured using the PMX instrument (RTI Electronics). The screen-film absolute sensitivity (\(K_{c,\text{air}}\) needed to form net optical density 1.0) was measured behind 20 mm Al additional filter. Measurements were performed at both 70 kV and 140 kV.

2.4. Image digitisation, calibration and transfer
The images were digitised to 12 bits with 170 µm pixel size (150 dpi) using a Vidar VRX-12 digitiser at Malmö University Hospital (partner 2). A square root translation table was selected for the digitisation and for the lumbar spine images a linear translation table was also used. A film step wedge, with known diffuse optical densities, was digitised for calibration of the digitiser. The images (4-10 MB each) were transferred to the computers in Linköping and London through use of the Internet.

Some of the lumbar spine images were exposed using a contrast equalisation filter between the x-ray tube and the KAP-meter. These patient images were not included in the image analysis below.

2.5. Image analysis
2.5.1. Discussions with radiologists
Local radiologists at both LKP and RMH were contacted and image features, details and characteristics discussed. The CEC Image Quality Criteria Document and VaOU et al. (1995), which provides criticisms of the previous text for chest images, were used as a basis for these discussions.

2.5.1.1. Chest Images
At each centre, the local radiologist concentrated on different aspects of interpreting chest images. LKP discussions concentrated on the general anatomy of the chest regions whereas, RMH talks were more focused on the image details. Both normal and AMBER films were viewed.

The vascular pattern of the lung could be viewed to the lung edges and gave evidence of inflammation. Good imaging of the bronchial tree, borders of the heart, aorta, diaphragm and costophrenic angles was more related to the individual anatomy and projection than to the unsharpness of the imaging system and viewing the bronchial tree showed the importance of dynamic range. Details within the retrocardiac lung and mediastinum were difficult to see without employing Amber.
Calcifications and the line (fissure) across the lung in the PA view, due to the lung lobes, are classified as high contrast details. Blood vessels and nodules in the lung are classified as low contrast details and have similar attenuation. Detection of low contrast details depends on their position in the lung and their sharpness can be lost due to respiration. End on vessels in the direction of the beam have higher attenuation. Reticular details are air filled cavities in the lung. Overall, the requirements specified by the CEC document appear difficult to meet, especially for the LAT view, and the recommended detail sizes seem too small.

From these discussions, it was jointly decided to measure contrast and detail size from profiles of blood vessels in the right costophrenic angles, retrocardiac region to the left of the spinal column, and the central right lung away from its periphery and apices in the PA projection. For the LAT projection, measurements in the costophrenic angles, retrocardiac region and the lung region above the retrocardiac area were to be undertaken. A number of large and small vessels in each of these three areas, corresponding to low, high and medium optical density respectively, were to be measured.

2.5.1.2. Lumbar Spine Images

Lumbar spine images were discussed with local radiologists on a separate occasion. Normal radiographs were again employed to aid the discussion. The meetings covered similar topics for both LKP and RMH.

The sacro-iliac joints can clearly be seen on PA/AP radiographs depending on field alignment though both may not be in the image at the same time. Assessment of its curvature and texture may be useful in determining pathology. Disk spaces, end plates, pedicles and spinous processes were clearly reproduced in the PA/AP projection and the last three had high contrast due to their cortical bone content. The transverse processes in the PA/AP view and spinous processes in the LAT projection had lower contrast. The end plates in the LAT view were less visible than for the PA/AP view.

Cortex and trabecular structures were visible on the vertebra and are taken as high contrast details. The trabecular details have similar sizes to those stated in the CEC document. The psoas shadow (visible in the image due to its surrounding adipose tissue layer) in the PA/AP view on either side of the spinal column is classified as a low contrast detail. Detectability of the anatomy in both views can be impaired by stomach and bowel contents.

From these discussions, it was jointly decided to take measurements of contrast from profiles over the edges of the transverse processes in the PA/AP view and the spinous processes in the LAT view, and across the sacro-iliac joint and the psoas muscle in the PA/AP projection.

2.5.2. Measurements in digitised images

2.5.2.1. Contrast

Measurements in the digitised images were made using the software IMAGE TOOL (LKP) and ANALYZE (RMH). The contrast of important image details in chest and lumbar spine images was measured by allowing profiles across details of interest to be registered. The pixel values were converted to optical densities and to air-kerma at the
film by use of the digitiser calibration and the characteristic curve of the film. A polynomial of the 7th-9th order was fitted to the pixel data points and used to calculate the contrast of selected details. The contrast was calculated as

\[ C = \frac{K_{\text{max}} - K_{\text{min}}}{K_{\text{max}}} \]  

(1)

where \( K_{\text{max}} \) and \( K_{\text{min}} \) are the air kerma at the film beside and behind the contrasting detail, respectively.

The size of the detail (blood vessel) was characterised by the full width half maximum (FWHM) distance. In the chest PA images, three different areas were investigated: the lung area, the costophrenic angle area and retrocardiac area. For each area 2-5 profiles were selected. In the lumbar spine images (PA view), the contrast of the transverse processes and the contrast of the psoas muscle were measured as well as the contrast of the sacro-iliac joint. In the LAT view, the contrast of the spinous processes was measured.

**2.5.2.2. Dynamic range**

The dynamic range in both examinations was measured. Due to the limited dynamic range of the Vidar VRX-12 digitiser, and its comparably large noise at high optical densities, measurements using the pixel value frequency distribution (histogram) were complemented with direct measurement of maximum and minimum optical densities with a densitometer. The dynamic range was calculated as the difference between the 1% and 99% points of the cumulative air-kerma or optical density distribution.

The dynamic range of the films was characterised as the quotient of the air-kerma values at the film (corresponding to a given optical density, OD) for which the film gradient (film-\( \gamma \)) exceeded the value 0.5.

**2.5.2.3. Noise**

It is impossible to make a precise measurement of the noise on a clinical image: this would require a priori knowledge of the imaged object. One approach, however, which has some utility, is to estimate the noise by subtracting a noise suppressed image from the original image. We have used a median filter to calculate the noise suppressed image and a code has been written to median filter the image and to make the subtraction. The resulting median filtered and subtracted images were written in the ANALYZE image format so that they could be viewed. The median filter works very well for a constant background and preserves edges in the image. Structures which are smaller than the filter size are removed and appear in the noise. The filter size was therefore set to 3x3. Studies with 5x5 and 7x7 filters show that this was a reasonable choice.

A study was made of the variation of the noise level at fixed optical density from image to image. Noise in 15 image patches was calculated for both chest PA and lumbar spine PA/AP views. The patches were selected to cover a range of image features and optical densities. The pixels in each patch were divided into sets with the same gray level and the rms value of the noise at each gray level was calculated. Gray level values occurring in just a few pixels were rejected. The resulting noise values were converted from gray level to optical density values by multiplying with the gradient of the function which relates these two quantities. The results at fixed optical density values were plotted against optical density.
density for the different patches were compared with each other and with the digitiser noise supplied by PTB. Because of the large contribution to the measured noise values from the digitiser noise, no measurements were made on the LAT view radiographs.

3. Results and discussion

3.1. Patient absorbed dose

The energy imparted to the patients in the PA and LAT views as function of the entrance surface dose is shown in figure 1. The median entrance surface dose, energy imparted and mean absorbed dose are given in Table 1.

![Figure 1](image.png)

**Figure 1.** The energy imparted to the patients as a function of the entrance surface dose, ESD in (a) chest PA and (b) chest LAT, (c) lumbar spine PA and (d) lumbar spine LAT views.

<table>
<thead>
<tr>
<th></th>
<th>Chest (n=29)</th>
<th>Lumbar Spine (n=21)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PA LAT</td>
<td>PA LAT</td>
</tr>
<tr>
<td>Entrance surface dose (mGy)</td>
<td>0.20 ± 0.06</td>
<td>0.64 ± 0.67</td>
</tr>
<tr>
<td>Energy imparted (mJ)</td>
<td>1.6 ± 0.5</td>
<td>3.7 ± 4.1</td>
</tr>
<tr>
<td>Mean absorbed dose (µGy)</td>
<td>0.021 ± 0.06</td>
<td>0.048 ± 0.05</td>
</tr>
</tbody>
</table>
The median entrance surface dose from this survey of chest and lumbar spine are between 33-64% lower than the entrance surface doses for a standard-size patient given by the European Commission (1996).

3.2. Imaging system characteristics
The imaging system characteristics are given in Table 2 below. The chest stand uses automatic exposure control but the lumbar spine does not which may cause some additional variation in optical densities. The HVL is comparably low for a chest stand. A stationary grid with strip frequency 40 cm⁻¹ was used in the chest stand which results in that the lead strips are present in the images; in the lumbar spine the grid is moving and the lead strips are therefore invisible.

Table 2. Imaging system characteristics.

<table>
<thead>
<tr>
<th></th>
<th>Chest</th>
<th>Lumbar Spine</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-ray generator</td>
<td>CGR Phasix 65</td>
<td>Siemens</td>
</tr>
<tr>
<td>Focal Spot size (mm)</td>
<td>0.6/1.2</td>
<td>0.6/1.2</td>
</tr>
<tr>
<td>Automatic Control</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Tube Potential (kV)</td>
<td>140</td>
<td>70-80</td>
</tr>
<tr>
<td>Half-value layer (mm Al)</td>
<td>4.45 (at 140 kV)</td>
<td>2.97 (at 70 kV)</td>
</tr>
<tr>
<td>Focus-film distance (mm)</td>
<td>1432</td>
<td>1100</td>
</tr>
<tr>
<td>Anti-scatter grid (N, r, d)</td>
<td>40 cm⁻¹, ratio=12</td>
<td>70 cm⁻¹, ratio=16, 36 µm</td>
</tr>
<tr>
<td>Grid covers and interspaces material</td>
<td>Al</td>
<td>Al</td>
</tr>
<tr>
<td>Fluorescent screen (Gd₂O₂S)</td>
<td>Agfa Curix Ortho Med.</td>
<td>Kodak Lanex Med.</td>
</tr>
<tr>
<td>Film</td>
<td>Fuji Super HRL</td>
<td>Fuji Super HRE</td>
</tr>
<tr>
<td>Cassette front material</td>
<td>2.5 mm ABS plastic + 1.6 mm Al +</td>
<td>0.25 mm vinyl</td>
</tr>
<tr>
<td>and thickness</td>
<td>50 µm steel</td>
<td>240</td>
</tr>
<tr>
<td>Measured Speed</td>
<td>240</td>
<td>340</td>
</tr>
<tr>
<td>Film developer</td>
<td>Agfa Curix HT 530</td>
<td>Kodak RP X-omat</td>
</tr>
</tbody>
</table>

These system parameters were compared to the examples of good radiographic technique given by the European Commission (1996). For the chest examination, the measured HVL and the speed of screen-film system is lower than those recommended by the European Commission. In the lumbar spine examination, the tube potential used is lower than those recommended by the European Commission.

3.3. Image analysis
3.3.1. Low-contrast details
3.3.1.1. Chest
An example of a profile across a blood vessel in the lung area is shown in figure 2. The results from the contrast measurements are displayed in figure 3 (PA) and 4 (LAT) where the size (FWHM) is plotted as function of the contrast in all three areas.
Figure 2. Profile across a blood vessel in the lung area. The dots are the pixel values and the solid line a 9th order polynomial fitted to the pixel values. The contrast and size (FWHM) of the vessel were measured to 13% and 1.6 mm, respectively.

Figure 3. The distribution of size (FWHM) and contrast of blood vessels in the chest PA radiographs in each of the three areas of interest; (a) lung area, (b) costophrenic angles and (c) retrocardiac area. The solid lines are fitted to the measured values using linear regression.

Figure 4. The distribution of size (FWHM) and contrast of blood vessels in the chest LAT radiographs in each of the three areas of interest; (a) lung area, (b) costophrenic angles and (c) retrocardiac area. The solid lines are fitted to the measured values using linear regression.
Table 3. Results of contrast and size (FWHM) measurements in chest radiographs. The table gives the mean values ± 1 relative standard deviation of the mean.

<table>
<thead>
<tr>
<th></th>
<th>Chest PA</th>
<th>Chest LAT</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Contrast (%)</td>
<td>Size (mm)</td>
</tr>
<tr>
<td>Lung area</td>
<td>13.1±4.7</td>
<td>2.2±0.7</td>
</tr>
<tr>
<td>Costophrenic angle area</td>
<td>12.3±4.8</td>
<td>2.3±0.9</td>
</tr>
<tr>
<td>Retrocardiac area</td>
<td>14.2±5.2</td>
<td>2.8±1.0</td>
</tr>
</tbody>
</table>

The average contrast and average size (FWHM) are not directly connected since they are averages of a large number of vessels.

The vessel size was characterised by the FWHM in the image. What we really want to estimate is the size of the vessel in the patient. This required some correction factors. These are

- imaging system magnifications;
- correlation between FWHM and vessel diameter;
- influence of vessel orientation with respect to the image plane.

The magnification in the radiograph is between 0 and 25% with an estimated mean of 9% given that the average thickness of the patient is 24 cm and FFD=143.2 cm; \((143.2\,\text{cm}/(131.2-24/2)\,\text{cm}=1.09)\).

It could be argued that the diameter of the vessel would be approximately 2xFWHM.

The orientation of the vessel will not always be parallel to the image plane but vary between 0 and 90 degrees. Vessels that were oriented perpendicular to the image plane (90 degrees) have not have been measured. It was estimated that the maximum orientation of any measured vessel would be 70 degrees. The additional pathlength through the vessel will then be the mean of \((1/\cos\theta)\) integrated from 0 to 70 degrees where \(\theta\) is the angle between the vessel and the image plane. This factor was calculated to 1.4.

The relation between the actual thickness of the vessel in the patient, \(d_{\text{pat}}\) based on measurement of the FWHM in the image was calculated from.

\[
d_{\text{pat}} = \text{FWHM} \cdot 2 \cdot 1.4/1.09 = 2.57 \, \text{FWHM}
\]

3.3.1.2. Lumbar spine

Results of measurement of the contrasts of the transverse and spinous processes as well as the psoas muscle and the sacro-iliac joint are given in table 4. Sample histograms of the results are shown in figure 5.
Table 4. Results of contrast measurement in lumbar spine radiographs. The table gives the mean values ± 1 relative standard deviation of the mean.

<table>
<thead>
<tr>
<th></th>
<th>Contrast (%) ± 1 SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transverse processes</td>
<td>8.9 ± 5.3</td>
</tr>
<tr>
<td>Spinous processes</td>
<td>14.3 ± 5.8</td>
</tr>
<tr>
<td>Psoas muscle shadow</td>
<td>8.8 ± 4.9</td>
</tr>
<tr>
<td>Sacro-Iliac joint</td>
<td>36.9 ± 15.6</td>
</tr>
</tbody>
</table>

Figure 5. Distribution of contrast values obtained for (a) the transverse processes in the lumbar spine PA view and (b) spinous processes in the lumbar spine LAT view.

3.3.2. High contrast detail objects

It was found that the spatial resolution of the digitiser was not sufficient (170 µm) and that interpretation of measurements on small high contrast details would be very difficult. The selection of these details was therefore made based on our discussion with local radiologists and using the CEC Image Quality Criteria. In the chest, 0.3-0.5 mm calcifications in the lung apices will be considered as a small, high-contrast object. In the lumbar spine, trabecular structures of sizes 0.3 and 0.5 mm will be used.

3.3.3. Dynamic range

The dynamic range of the chest and lumbar spine radiographs are given in Table 5. Figure 6 shows the dynamic range (expressed in optical density or air kerma) as function of patient weight for the chest and lumbar spine PA and LAT views, respectively.

Table 5. The range of optical densities observed in the chest and lumbar spine radiographs and the corresponding range of $K_{c,air}$ at the screen-film system. The table gives the mean values ± 1 relative standard deviation of the mean.

<table>
<thead>
<tr>
<th></th>
<th>Chest PA</th>
<th>Chest LAT</th>
<th>Lumbar Spine PA</th>
<th>Lumbar Spine LAT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic range, (ODU)</td>
<td>1.46 ± 0.15</td>
<td>1.68 ± 0.26</td>
<td>1.93 ± 0.67</td>
<td>2.66 ± 0.61</td>
</tr>
<tr>
<td>Dynamic range, $K_{c,air}$ (µGy)</td>
<td>7.1 ± 1.0</td>
<td>11.0 ± 8.9</td>
<td>6.5 ± 4.1</td>
<td>11.5 ± 4.7</td>
</tr>
</tbody>
</table>
Figure 6. The dynamic range as function of patient weight in the chest (a-d) and the lumbar spine (e-h). The dynamic range is expressed either in terms of optical density or in air kerma at the screen-film system.
The dynamic range of the films are given in table 6.

**Table 6. Dynamic range of the films used in chest and lumbar spine radiographs.**

<table>
<thead>
<tr>
<th></th>
<th>Chest (Fuji Super HRL)</th>
<th>Lumbar spine (Fuji Super HRE)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air kerma (min, max) (µGy)</td>
<td>1.15, 36.3</td>
<td>0.74, 17.37</td>
</tr>
<tr>
<td>Air kerma difference (max-min) (µGy)</td>
<td>35.2</td>
<td>16.6</td>
</tr>
<tr>
<td>Air kerma ratio, max/min</td>
<td>31.6</td>
<td>23.5</td>
</tr>
</tbody>
</table>

3.3.4. Noise

Figure 7 shows the rms noise estimated from 15 patches for the chest PA view and 15 patches in the lumbar spine PA/AP view, respectively. Also shown is the measurement of the digitiser noise made by PTB. It will be seen that the noise is dominated by the digitiser noise. It is also seen that the results cluster well and that the noise can be predicted reasonably well on the basis of the optical density. Close examination of the subtracted images for the chest radiographs shows that the lines of the stationary grid are visible. The imaged grid thus appears as part of the noise. This is not important at present because of the large noise introduced by the digitiser, but would need to be taken into account in any analysis of data obtained using an instrument with higher performance. Analysis of radiographs taken without the use of a stationary grid would of course prevent this problem but it may well be that the presence of a stationary anti-scatter grid will reduce the performance of the imaging system in situations where the signal-to-noise ratio is low.

Although these results are useful and will enable us to make some validation of our noise calculations, the noise introduced by the digitiser is an important limitation, and as noted above, it was decided not to analye the noise levels for the other two projections. If higher quality digitised images become available early in 1997, this analysis may be repeated.

![Figure 7](image_url)

**Figure 7.** Estimated rms noise in (a) the chest PA view and (b) the lumbar spine PA/AP view.
3.4. Structures selected for inclusion in the Monte Carlo model

One of the aims of the work carried out in the first year of the project was to obtain a set of images from which features could be extracted and incorporated into the voxel Monte Carlo model. The aim has been achieved and Table 7 summarises those structures which will be used as part of the calibration of the model for chest and lumbar spine images in both PA and LAT examinations.

The selection of the high contrast details was discussed in section 2.5 above. The selection of the low contrast details is discussed below.

3.4.1 Chest Images

Blood vessels in the chest images were identified as low contrast details. Transverse processes and the psoas muscle for the PA view and spinous processes for the LAT view were selected as low contrast objects in lumbar spine images.

The size of the blood vessels in the chest images were determined from examining cumulative distributions of the registered values of the FWHM (Figs. 3 and 4) for the central lung, costophrenic angles and the retrocardiac regions, respectively in both views. It was decided to select the size corresponding to the values of FWHM for which 30% of the vessels had a lower FWHM. Small vessels are more difficult to detect and the selection of a small vessel size will provide more stringent imaging requirements for the modelling.

The contrast, \( C \), of the vessel was fitted to the FWHM, \( X \), (Figs. 3 and 4) by linear regression for each set of results.

\[
C = \alpha X \tag{3}
\]

where: \( \alpha \) is the linear regression fitted coefficient. The values of the coefficient are shown in Table 8 below.

<table>
<thead>
<tr>
<th>Region</th>
<th>Examination</th>
<th>PA</th>
<th>LAT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Lung</td>
<td>5.37 ± 0.25</td>
<td>4.16 ± 0.27</td>
<td></td>
</tr>
<tr>
<td>Costophrenic Angles</td>
<td>4.80 ± 0.24</td>
<td>3.46 ± 0.24</td>
<td></td>
</tr>
<tr>
<td>Retrocardiac</td>
<td>4.78 ± 0.22</td>
<td>3.80 ± 0.18</td>
<td></td>
</tr>
</tbody>
</table>

These coefficients were used to estimate the contrast of the vessels specified in Table 7a as shown in Table 9 below:
Table 9. Estimated contrast using the vessel size, FWHM, specified in Table 7 for different regions in PA and LAT chest examinations. The uncertainties correspond to the standard error of the fitted coefficient.

<table>
<thead>
<tr>
<th>Region</th>
<th>FWHM (mm)</th>
<th>Examination</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>PA</td>
</tr>
<tr>
<td>Central Lung</td>
<td>1.8</td>
<td>9.67 ± 0.45</td>
</tr>
<tr>
<td>Costophrenic</td>
<td>1.8</td>
<td>8.64 ± 0.43</td>
</tr>
<tr>
<td>Retrocardiac</td>
<td>2.0</td>
<td>9.56 ± 0.44</td>
</tr>
<tr>
<td>&quot;</td>
<td>3.0</td>
<td>14.34 ± 0.66</td>
</tr>
<tr>
<td>&quot;</td>
<td>4.0</td>
<td>19.12 ± 0.88</td>
</tr>
</tbody>
</table>

It should be appreciated that the results in Table 9 for the different regions may correspond to different angulations of the blood vessels. For example, the scatter to primary ratio in the retrocardiac space will be high, and to achieve the same contrast for a given vessel diameter, a steeper angulation is necessary. The true vessel diameter will be greater than the FWHM, as noted in section 3.3.1.1, depending on the angulation. In Eq. 2, the angulation was assumed to be fixed at an angle between 0° and 70°.

The composition of the blood vessels will be obtained from ICRU Report No. 46. Their thicknesses may need to be adjusted to calibrate the voxel Monte Carlo model, as the angulation of the vessels measured was unknown.

3.4.2 Lumbar Spine Images

The thicknesses of the transverse processes and the spinous processes were obtained from calliper and micrometer measurements of an actual skeleton at RMH. It was realised that these measurements may not be representative of the anatomy in the images though they may still provide a starting point from which the voxel Monte Carlo model can be adjusted.

The majority of the measurements were undertaken for processes on the L5 vertebra. The inclusion of the L1 vertebra is to facilitate results for alternative surrounding anatomy. The processes appeared compact and smooth which implied that they were composed of cortical bone whereas other vertebral bodies had trabecular structure (high contrast detail, see Table 7) which was produced by red bone marrow cavities in cortical bone.

The psoas shadow is a layer of fat between the psoas major and minor muscles. Contrast was produced by the difference in the attenuating properties of adipose tissue and muscle. The data set of contrast measurements is small since bowel contents impaired the detection of the psoas shadow in the majority of the images. The object was usually visualised close to the L2 transverse process. The fatty layer curved round between the two muscles. This meant that it was difficult to estimate the thickness of the detail at any particular position. Curvature with respect to the
incident radiation increased the x-ray pathlength. However, the size stated in Table 7 will provide a useful starting point for the simulations.

The average contrast results for the sacro-iliac joint have been reported in Table 4. This object will not be incorporated into the voxel Monte Carlo model as it was found that the contrast could vary greatly depending on the position in which line profiles were taken.
Table 7a. Structures and their positions selected for inclusion in the Monte Carlo voxel-phantom based model (Chest)

<table>
<thead>
<tr>
<th>Chest  PA</th>
<th>Low contrast details</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Lung</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 1.8 mm</td>
<td>Positions: Central right lung</td>
</tr>
<tr>
<td><strong>Costophrenic Angles</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 1.8 mm</td>
<td>Positions: 1 cm up and 1 cm in from the lowest visible point.</td>
</tr>
<tr>
<td><strong>Retrocardiac Region</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 2 mm, 3 mm and 4 mm</td>
<td>Positions: 50% up from the diaphragm and 25% across from the border with the left lung</td>
</tr>
<tr>
<td>Composition: The vessels are to be composed of blood with healthy lung as the background material. (ICRU 46 compositions)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

| High contrast details | Structure of interest: Calcification | Size (FWHM): 0.3 and 0.5 mm: spherical or disk shaped | Position: Lung apices |
| Composition: Calcium hydroxyapatite, 3.3 g.cm\(^{-3}\) |

<table>
<thead>
<tr>
<th>Chest LAT</th>
<th>Low contrast details</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Lung</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 1.8 mm</td>
<td>Positions: 2 cm up from top edge of heart, upper region of lung free from projection of major vessels and spine.</td>
</tr>
<tr>
<td><strong>Costophrenic Angles</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 1.8 mm</td>
<td>Position: Close to lowest visualised part of lung and avoiding the spine</td>
</tr>
<tr>
<td><strong>Retrocardiac Region</strong></td>
<td>Structure of interest: Blood vessel</td>
<td>Size (FWHM): 2 mm, 3 mm and 4 mm</td>
<td>Positions: 2 cm up from bottom edge</td>
</tr>
<tr>
<td>Composition: Same as above for PA view.</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

| High contrast details | Same as above for PA view. |
Table 7b. Structures and their positions selected for inclusion in the Monte Carlo voxel-phantom based model (Lumbar Spine)

**Lumbar spine PA**

**Low contrast objects**

<table>
<thead>
<tr>
<th>Transverse processes</th>
<th>Structure of interest: L1 and L5 vertebra</th>
<th>Size: 5 mm and 2 mm</th>
<th>Positions: Just beside the lumbar spine</th>
<th>Composition: Cortical bone within ICRU muscle</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Psoas muscle</strong></td>
<td>Structure of interest: 2-4 mm wide layer of fat</td>
<td>Size: with greater than a 4 mm pathlength</td>
<td>Positions: 2-3 cm away from the L2 vertebrae</td>
<td>Composition: Adipose tissue within ICRU muscle</td>
</tr>
</tbody>
</table>

**High contrast objects**

| Structure of interest: Trabecular structure | Size: 0.3 and 0.5 mm | Positions: In the vertebrae in the lumbar spine | Composition: Red bone marrow cavities in the compact bone (ICRU Report No. 46 composition) |

**Lumbar spine LAT**

**Low contrast objects**

<table>
<thead>
<tr>
<th>Spinous processes</th>
<th>Structure of interest: L1 and L5 vertebrae</th>
<th>Size: 6 mm and 3 mm</th>
<th>Positions: Just beside the lumbar spine</th>
<th>Composition: Cortical bone within ICRU muscle</th>
</tr>
</thead>
</table>

**High contrast objects** Same as above for PA view.
4. Summary and conclusions

The objective of this work is to develop a realistic voxel Monte Carlo model and to use it for the prediction of image quality and patient dose as a function of the technique parameters. The model will then be used to optimise the technique parameters such that sufficient image quality can be obtained at the lowest possible doses.

A set of radiographs have been acquired and digitised for software manipulation. Measures of contrast, dynamic range and noise have been extracted for features of diagnostic importance, in chest and lumbar spine images, selected after consultation with local radiologists. Detailed information was assembled on technique factors for the imaging systems used to produce the films. Patient specific information was also gathered for each radiograph including measures of entrance surface dose and kerma-area product converted to energy imparted to the patient.

The first stage of this work (acquisition of relevant data from clinical images) has been completed. The next stage is the calibration of the voxel Monte Carlo model utilising the technical, patient and anatomical data specified in Table 7. In the calibration, parameters such as tissue composition, tissue density and patient dimensions will be varied to match the results using the model with the image quality values and the patient dose values obtained in the reported measurements.

The aims of the first year of the project have been met and work will continue with the calibration of the model.

Acknowledgements

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References


