Radiation detectors in medical and biological applications

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Abstract

Basic definitions of imaging theory are given and their importance for detector development is demonstrated. Developments in two-dimensional and scanning radiographic systems are discussed. The substantial progress in non-invasive coronary angiography with synchrotron radiation is presented together with a comparison of the performance of a gaseous and a silicon detector. Synchrotron radiation is also applied in mammography. New modalities making use of coherence are discussed. Non-radiative methods competing with X-ray systems are also presented. Detectors for nuclear medicine and biological structure research are briefly mentioned. © 1998 Elsevier Science B.V. All rights reserved.

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1. Introduction

Since most of the medical and biological applications of radiation detectors are concerned with imaging some basic properties of imaging systems and the terminology of the imaging community will first be discussed. The main purpose of this part is to convince the reader that a comparison of the performance of different detectors is not straightforward and that for such a comparison appropriate methods have to be used.

Since there is now general agreement in medical imaging that the radiation dose has to be kept at the lowest possible value compatible with the diagnostic purpose, a medical image is always noise limited, at least in some region of interest. In the language of the imaging community Poisson statistics are included in the noise.

2. Characterisation of a noise limited imaging system

The noise properties of an imaging system can be determined from the noise autocorrelation function $C(x, y)$. This function is given by the correlation of the fluctuations $\Delta S$ of a constant (in the mean) signal as a function of the distance:

$$C(x, y) = \frac{1}{a} \int_{a} \int_{a} \Delta S(x', y') \Delta S(x' + x, y' + y) \, dx' \, dy'. \quad (1)$$

The variables $x, y, x', y'$ are (Cartesian) coordinates in the image and $a$ is its area. The Fourier transform
$W(f, g)$ of the noise autocorrelation function is Wieners noise power spectrum

$$W(f, g) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} C(x, y) e^{-2\pi i (f \cdot x + g \cdot y)} \, dx \, dy.$$  \hfill (2)

Here $f$ and $g$ are the spatial frequencies with dimensions of inverse length related to $x$ and $y$, respectively. Spatial frequencies are often quoted in line-pairs/mm. In the frequency domain it is possible to make use of the convolution theorem to simplify the analysis of composite systems. Consequently, instead of the point spread function $PSF(x, y)$ (i.e. the detector resolution curve) its Fourier transform, the optical transfer function $OTF(f, g)$ given by

$$OTF(f, g) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} PSF(x, y) e^{-2\pi i (f \cdot x + g \cdot y)} \, dx \, dy$$

or its modulus, the modulation transfer function $MTF(f, g)$ given by

$$MTF(f, g) = |OTF(f, g)|$$  \hfill (3)

is used. Note that $MTF(0,0) = 1$ and also that for a detector of infinitely good resolution $MTF(f, g) = 1$.

The total $MTF$ of a detector system is the product of the $MTF$s of its components. The use of the (spatial) frequency-domain permits in some cases an improvement in the spatial resolution by deconvolution, normally, at the expense of increased image noise.

The noise properties of an imaging detection system, its quantum efficiency and its spatial resolution are usually combined in the detective quantum efficiency $DQE(f, g)$

$$DQE(f, g) = \left( \frac{SNR_{out}(f, g)}{SNR_{in}(f, g)} \right)^2 = \left( \frac{S_{out}(f, g)\sigma_{in}(f, g)}{S_{in}(f, g)\sigma_{out}(f, g)} \right)^2,$$  \hfill (5)

where $SNR$, $S$ and $\sigma$ are the signal-to-noise ratio, the signal and the noise at the input and the output of the detector system, respectively.

Obviously, for a detector for which the noise is given by Poisson statistics only and which has infinitely good spatial resolution, the $DQE$ is equal to ordinary quantum efficiency $QE$. In general, the $DQE$ may be expressed by the functions already defined, giving

$$DQE(f, g) = QE \left( \frac{MTF(f, g)}{W(f, g)} \right)^2.$$  \hfill (6)

The ‘zero-frequency $DQE$’, $DQE(0,0)$, is used to describe the frequency-independent properties of an imaging system. It is again equal to ordinary $QE$ for a Poissonian sampling system of finite pixel size.

The following example gives some insight into the meaning of the zero-frequency $DQE$: for a one-dimensional integrating X-ray detector with a finite (intrinsic) single-event energy resolution, a resulting normalised spectrum $n(E)$ for the energy deposited per event and additional integrator noise $W_{add}$ expressed in equivalent photons, the zero-frequency $DQE$ is given by

$$DQE(0) = \frac{QE}{1 + \frac{QE}{\left( \frac{1}{E} \right)^2 n(E) E^2 dE - E^2} + \frac{W_{add}}{QE N}},$$  \hfill (7)

where $N$ is the number of photons accumulated and $E_i$ the photon energy. The central term in the denominator describes the loss of signal-to-noise ratio due to, e.g. escape losses from fluorescence or a Compton contribution. This loss can never be recovered by improving statistics. The additive noise term, on the other hand, disappears with increasing statistics. Therefore, $DQE(0)$ depends on the signal size and approaches $QE$ for large signals and a narrow spectrum $n(E)$. The resulting curves are shown in Fig. 1 for different noise levels. Also, shown are $DQE$ curves for a counting system, where $DQE$ approaches zero for a rate slightly above the inverse deadtime.

If the pixel noise in a detector is measured to determine $QE$, naively using Poisson statistics, easily an apparent $QE > 1$ can be obtained. This is most likely the case for a low-noise sampling system with some ‘cross talk’ in the detection system (wide presampling $PSF$), because a pixel would ‘see’ part of the statistics of its neighbours. Therefore, even for this simple task the methods described above have to be used. For further information the
Fig. 1. Zero-frequency DQE as a function of the number of photons per pixel at the input of the detector for QE = 0.7. Curves (a–d): integrating systems with a noise of 10, 10^2, 10^3 and 10^4 equivalent photons, curves (e–i): counting systems with a ratio of exposure time to deadtime of 10^3, 10^4, 10^5, 10^6 and 10^7.

As a consequence, there is only one reliable method to compare the performance of different detectors, namely to compare DQE as a function of the spatial frequencies at the most critical intensity.

3. Digital radiography

The by far most common application of medical imaging is radiography, and to a lesser extent, fluoroscopy. The first is still largely dominated by X-ray film and screen film systems as detectors, but industry has made a strong effort to replace them by digital systems with all the possibilities of processing and transferring digital data. These systems, if well made, have the additional advantage of a much larger dynamic range compared to the seven bit of film. Therefore, by playing with the contrast function, the medical doctor can see details which otherwise would require several exposures, even for digitised film data. Present developments go into two directions: two-dimensional general purpose systems and scanning systems.

3.1. Two-dimensional systems

The quantum efficiency of a 300 μm Si-wafer for X-rays of medical energies is at best a few percent. Therefore, since QE is an upper bound for DQE, the idea of using standard Si-wafers as 2D-detectors can immediately be abandoned.

With the advent of the amorphous silicon (a-Si) thin-film technique, detectors based on a converter and an a-Si active matrix became an attractive possibility, because they can be built at a reasonable price and in sizes and weights compatible with common X-ray equipment. Ideally, they would fit into a standard X-ray cassette. Such systems are at the industrial production threshold and will soon be produced in large quantities.

There are two types of such systems using indirect or direct conversion of the X-ray photons. The indirect systems use a scintillator as converter and an active matrix as a detector for the visible photons emitted by the scintillator. The direct systems use a heavy semi-conductor as a converter with direct charge transfer to the active matrix. The charges stored in the capacitance of the pixels of the reader is referred to Refs. [1–4]. In these references the effects of spatial sampling are also discussed.

The noise properties of an imaging system, obviously, determine the radiation dose which a patient has to accept for a given signal-to-noise ratio. The fraction of the radiation dose a patient has to take due to imperfections of the imaging system is \( \frac{1}{DQE} \). Therefore, for medical applications DQE should be close to one.

For a diagnostic interpretation a signal-to-noise ratio of three to five is the minimum required. For a sampling system with matched pixel size the skin dose \( D_{\text{skin}} \) increases with the inverse fourth power \( (1/w^4) \) for a cubic object of size \( w \), and also increases with the square of the signal to noise ratio \( SNR_{\text{AS}} \) required [5]. \( D_{\text{skin}} \) is given by

\[
D_{\text{skin}} = \frac{2e^\alpha \cdot L \cdot SNR_{\text{AS}}^2}{DQE(0) \cdot \mu^2 w^4 C_m^2 \cdot E(y) \left( \frac{\mu}{\rho} \right)}. \tag{8}
\]

Here \( L \) is the length of the tissue traversed by the beam, \( \mu \) is the absorption coefficient of the tissue and \( \rho \) is its density. \( C_m = \Delta \mu / \bar{\mu} \) is the object contrast, i.e. the relative change of the absorption coefficient in the object with respect to its neighbourhood.

Any dose quoted without proper consideration of the spatial and intensity resolution is therefore meaningless.
active matrix are switched by thin-film transistors (TFTs) or thin-film diodes (TFDs) line by line to the output bus lines which follow the columns. The charges are transferred to an ADC via multiplexers, digitised and stored.

One such system (see Fig. 2), using a CsI-scintillator of 450 \( \mu \text{m} \) thickness, grown in needles of 6–10 \( \mu \text{m} \) in diameter, has been developed by Siemens and Thomson [6]. Its properties are summarised in Table 1. The readout speed of 12.5 frames/s makes the system also suitable for fluoroscopy.

A direct conversion system with a 2 mm amorphous Se converter is, e.g., under development in Saskatoon and in Toronto [7,8]. Images taken with this system can be found in Ref. [8].

Considering the present status of industrial development, it does not seem promising to start a new project in this field.

### 3.2. Scanning systems

Scanning systems (see Fig. 3) have two advantages compared to 2D-systems: if properly designed, they have reduced scatter background and they use only a small number of channels. But they need special scanning X-ray equipment and do not permit fluoroscopy.

The use of counting or integrating detectors is not a question of principle, as can be seen from Fig. 1. In integrating systems the spatial resolution is (de facto) always limited by the pixel size, while in counting systems an interpolation is possible. But integrating systems are always as good as counting systems in intensity resolution, if the integrator noise \( W_{\text{add}} \) (see Eq. (7)) is smaller than the Poisson noise. The improvement in cheap low-noise electronics has continuously decreased the limit for the use of integrating systems. In addition, integrating gaseous detector systems are robust, even with fine collecting structures.

The Novosibirsk commercial scanning system has been presented at this conference several times [9]. It uses a tapered pressurised gaseous detector originally working in the counting mode. With the transition to an integrating mode it follows the above mentioned line of development.

New developments use edge-on tapered Si-strip detectors (see Fig. 4 and Fig. 11). With this technique \( QE \) and \( DQE \) can be kept close to one due to the long active length. In addition, the well-known Si-technique can be used. These detectors work mainly in the counting mode.

A system of this kind has been developed in Trieste for application in digital mammography.
Table 2
Properties of the LBL-Berkeley Si edge-on strip detector

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness</td>
<td>300 ( \mu )m</td>
</tr>
<tr>
<td>Depth</td>
<td>45 mm</td>
</tr>
<tr>
<td>Strip pitch</td>
<td>100 ( \mu )m</td>
</tr>
<tr>
<td>Count rate</td>
<td>&gt; 1 MHz/segment</td>
</tr>
<tr>
<td>Segment in depth</td>
<td>2, 6, 12, 24 mm</td>
</tr>
<tr>
<td></td>
<td>tapered geometry</td>
</tr>
</tbody>
</table>

with synchrotron radiation [10,41,11] and is described in Section 4.3.

At LBL-Berkeley [12,42] a similar system has been developed, specially adopted to radiography with X-ray tubes. Its properties are shown in Table 2. The segmentation in depth permits a statistical energy measurement and an increase in the total rate. Fig. 5 shows MTF curves normal to the scan direction for this detector: a theoretical MTF derived from the pixel size and the results of an edge scan and a phantom measurement. Due to the very short track of the photoelectrons in Si (as opposed to gas), the MTF is mainly determined by the pixel size. Using a very narrow collimated beam the MTF in the direction of the scan is similar to Fig. 5 in spite of the wafer thickness of 300 \( \mu \)m. Fig. 6 shows the image of a mouse skull taken with this detector demonstrating its excellent resolution.

At Strasbourg work on a similar project is going on using \( \mu \)-strips of 50 mm depth and VLSI electronics. The main emphasis is on continuous readout, spectroscopic acquisition [13].

A different and promising approach is the use of heavy semiconductors (e.g. GaAs or CdTe) as line detectors. As an example, the fast counting GaAs
system developed by a Cagliari/Catania group and presented at this conference [14] is mentioned.

4. Synchrotron radiation in coronary angiography and mammography

Synchrotron radiation in medical imaging is regularly used in intravenous coronary angiography and in several other projects in the development phase. All these projects make use of at least one characteristic property of synchrotron radiation, e.g. high power per energy bandwidth, small beam divergence, good stability and even coherence. Monochromators are used to provide narrow band beams.

4.1. Coronary angiography

At DESY the NIKOS system for coronary angiography [15,16] has been handed over to the medical doctors for evaluation [17]. Now, three to four patients are examined per day, with a total of 276 patients examined at the time of the conference. The system works with the now well-known method of K-edge subtraction. Iodine contrast agent is injected into a peripheral vein. Fig. 7 shows a side view of the system with the patient being moved through the beam. As detector a two line high-pressure KrCO$_2$ ionisation chamber is used with a pixel size of 400 $\mu$m and an active width of 150 mm.

To avoid motion blurring, images have to be taken in less than 10 ms for a 2D image and in less than 250 ms for a line scan system resulting in an exposure time of $< 1$ ms per line. The number of photons per pixel accumulated in the detector in 1 ms amounts to a few $10^2$ behind a partially iodine filled ventricle and to about $10^6$ behind the lung corresponding to about $10^9$ Hz. As can be seen from Fig. 1 or Eq. (7), this requires in case of integrating systems a noise as low as a few photons to keep $DQE$ close to $QE$ at the lower end of the range.

Consequently, the system has to have a dynamic range (maximum signal/integrator noise) of 18–19 bit. A counting system with a maximum rate close to $10^9$ Hz would need a very large number of segments in the beam direction, making such a system technically difficult and very expensive. Therefore, all systems developed for coronary angiography now use integrating electronics with more than 16 bit dynamic range and an integrator noise of a few photons.

The right coronary artery (RCA) and the left anterior descending branch (LAD) are relatively easy to image, while the left circumflex branch (LCX) is always partially obscured by the left

Fig. 7. Side view of coronary angiography systems with synchrotron radiation.
ventricle. Fortunately, the LCX has a 'large' diameter of 3–4 mm in this region. Therefore it can be visualised making use of structure dependent image processing. The benefit of image processing in noise limited images should not be overestimated, because any improvement in intensity resolution is at the expense of spatial resolution and vice versa. Clearly, the subtraction of large (average) structures can help for the interpretation, as does the K-edge subtraction method in spite of a factor of $\sqrt{2}$ lost in the intensity resolution for constant skin dose.

Fig. 8 shows two simultaneously taken images, one below (E1) and one above (E2) the K-edge, and the resulting subtraction image, demonstrating how the right coronary artery emerges from the background by the subtraction method. The stent (support mesh) in the proximal part of the RCA is made visible by deliberately unbalancing the weighting factors for the subtraction.

The skin dose is about 40 mSv per subtraction image. The total skin dose for three scans in two positions and a few low rate adjustment scans amounts to about 300 mSv, somewhat below the dose for a standard invasive coronary angiography.

A detailed discussion about performance and present status of the NIKOS system developed at DESY and in Siegen can be found in Ref. [18].
The medical evaluation should be finished by the end of this year. Then it will be decided, if this technique will be implemented as standard medical routine.

To give an idea about the time scale and the effort needed to arrive at this level, it is pointed out that a group of about 10 people (including medical doctors) for 12 years was required. More than half of them have worked full time on the project.

At the NSLS in Brookhaven/USA a similar system has been developed and about 30 patients have been examined with this system [19]. It uses, as detector, a Si(Li) bar of 5 mm depletion depth, carrying two lines of pixels. Five mm of Si is equivalent to ~ 1 absorption length at the Iodine K-edge of 33.17 keV. A comparison of the NIKOS and the NSLS system is presented in Section 4.2.

At the ESRF a cooled Ge two line pixel detector with 2 mm depth and 350 µm pixel size has been developed and is presently tested [20]. It uses 16 bit integrating electronics with additional switchable gain. It has not yet been used for patient studies.

Synchrotron radiation has also been tried at NSLS Brookhaven for bronchiography using Xenon as a contrast agent to measure lung ventilation. The method used is similar to coronary angiography but works at the Xenon K-edge. The contrast achieved is only moderate due to the low Xe density. Because xenon is a narcotic, it is difficult to develop a safe procedure.

4.2. Comparison of a Si(Li) and a gaseous detector for coronary angiography

A careful comparison of the two detection systems (NIKOS and the NSLS detector) has been performed [3] using the methods described in Section 2.

Fig. 9a and b show the MTF and Wiener noise power spectrum \(W\) (denoted by NPS in the figure) for the ionisation chamber. Since the square of the MTF drops faster with frequency than the noise, \(DQE\) decreases with increasing frequency. This behaviour is shown in Fig. 10a and b for the two detectors. As can be seen from the figure, \(DQE\) is better for the ionisation chamber than for the Si(Li) detector at signals of more than 20000 photons, typical for a free standing artery. At smaller signals the Si(Li) detector had a better \(DQE\), but its dynamic range was insufficient. As a consequence of this analysis both detectors were equipped with new electronics, improving mainly the noise for the ionisation chamber [18] and mainly the dynamic range and minimum exposure time of the Si(Li) detector. A dynamic range of nearly 19 bit and a noise of three photons has been achieved for the ionisation chamber. The \(DQE\) of the Si(Li) detector will always stay below 0.5 due to its insufficient X-ray absorption. It remains to remark that after some initial struggle the gaseous ionisation chamber turned out to be more reliable than the Si(Li) detector.
4.3. Mammography

From Eq. (8) it can be worked out that for a 30 μm in diameter micro-calcification, typical for a very early stage of breast cancer, and a SNR of (only) three, the resulting skin dose would be 50 mSv and the mean glandular dose about 20 mSv, even for an ideal detector with \( DQE = 1 \). This is an unacceptable high dose for a screening procedure. Therefore, especially in mammography, methods which improve the object contrast are of great importance. Two possible candidates will be described in Section 5.

Synchrotron radiation is used for mammographic studies mainly because a reduction of the radiation dose or a corresponding improvement in image quality can be expected. This improvement is caused by an optimised narrow band beam and scatter background reduction due to the line scan method used. Most advanced in this field is a collaboration working at the medical beam line SYRMEP at ELETTRA in Trieste [10,41].

The detector used is an edge-on Si strip detector (see Fig. 11) coupled to VLSI preamplifiers, discriminators and scalers developed in Strasbourg [11]. The mechanical layout is such that detectors may be extended sideways and stacked. The basic properties of the detector are listed in Table 3.

Fig. 12 shows an image of an excised breast taken with a narrow band beam at 18 keV with 100 μm step size. The mean glandular dose was 0.2 mSv only, matched to the step size and to the relatively large vascular calcification visible in the frame.

5. New modalities

It is also possible to make use of the relatively good coherence of mono-energetic synchrotron radiation beams to increase the object contrast. These methods are phase contrast imaging [21] and diffraction enhanced imaging (DEI) [22]. Both methods make use of the phase shift \( \phi \) of an electromagnetic wave travelling in \( z \) direction through an object with electron density \( \rho_e \) (see Fig. 13). The phase shift is given in lowest order by

\[
\phi(x,y) = - r_e \lambda \int \rho_e(x,y,z) \, dz, \tag{9}
\]

where \( r_e \) is the classical electron radius and \( \lambda \) the wavelength. The direction of the beam is denoted...
by $z$. If the phase shift depends on one of the transverse variables, e.g. $x$, the wave is refracted by an angle $\Delta z$ given by

$$\Delta z \approx \frac{\lambda}{2\pi} \frac{\partial \varphi(x)}{\partial x}$$  \hspace{1cm} (10)$$

In phase contrast imaging the detector is simply placed at a substantial distance (up to a few meters) downstream of the object, instead of directly behind it, as it is in most cases optimal for an absorption image. Phase contrast imaging is most effective for small objects.

From Eq. (9) a very faint image at first could be expected, because the phase shift is proportional to $\rho_e$ and therefore proportional to the nuclear charge $Z$, similar to the Compton cross section. Compton images are known to have a low contrast for this reason. The photo cross section, the main contribution to absorption images, on the other hand, is proportional to $Z^5$ apparently providing a much larger object contrast. But the expectation of a faint image is not justified for three reasons: (i) photo and Compton cross sections are proportional to $r_e^2$ instead of the much larger factor $r_e \lambda$ appearing in Eq. (9), (ii) the phase shift at a given energy is only weakly coupled to absorption (by a dispersion relation) and (iii) the interference with the primary wave gives a strong enhancement.

Fig. 14 shows an absorption (top) and a phase contrast (bottom) image of a bee taken at the SYRMEP beam line at ELETTRA [23]. The detector was a high resolution X-ray film. The increase in contrast is impressive.

Techniques making use of Eq. (10), like DEI, will give a strong edge enhancement due to the derivative in the equation. This can be very useful for the interpretation of images.
In DEI a diffracting analyser crystal is placed between object and detector and two images are taken, one on either side of the rocking curve [22] of the analyser crystal at the point of the steepest slope. For a good analyser the width of the rocking curve is in the micro-radian range. Thus, the coherently or incoherently scattered part of the beam does not appear on the detector, and the image is essentially determined by the refracted part following Eq. (10). For the formation of the image the scattered parts are as important as the absorbed part of the beam, thus substantially increasing the object contrast.

By taking the sum and the difference of the two images weighted with the slope of the rocking curve, a refraction angle image and an extinction (or apparent absorption) image is obtained. These images are shown in Fig. 15 for a mammographic phantom together with an ordinary absorption image [22]. The contrast in the extinction image is more than 25 times better compared to the absorption image. The very clear appearance of the fibrils in the upper right corner of the angle image demonstrates that additional information is available.
which cannot be extracted from the absorption or the extinction image alone.

The increased contrast in angle and extinction images makes the method of DEI a very promising candidate for mammography.

The usefulness of phase contrast imaging for mammography still has to be proven.

6. Non-radiative methods

In some fields, methods which do not use ionising radiation compete with X-ray imaging. The most common of these non-radiative methods are ultrasound techniques and magnetic resonance imaging (MRI). Both methods have a reputation of being non-invasive, but do not allow the imaging of small, rapidly moving structures like coronary arteries. Ultrasound, using Doppler techniques, permits also some functional imaging, e.g. flow measurements. This is the case for MRI only in very special cases. Two examples will now be discussed.

6.1. Intra-coronary ultrasound

The technique of intra-coronary ultrasound [24,25], contrary to the synchrotron radiation method described in Section 4.1 and to common ultrasound techniques, is an invasive method using a catheter introduced through a femoral artery, similar to ordinary selective coronary angiography. It is technically interesting, because it contains in a 3.5 F (≈ 1.2 mm diameter) catheter tip of 12 mm length a complete 64 element phase array sonar system including control electronics and space for a guide wire. The system works at 20 MHz, permitting a sub-100 μm resolution. It produces high-quality cross sections through the coronary arteries, because the transducer moves with the artery, thus overcoming the problem of motion blurring.

In the Doppler mode the system also permits blood flow measurements in arteries down to about 2 mm in diameter.

The technique is a good example for the progress allowed by micro-electronics.

6.2. Hyper-polarised $^3$He magnetic resonance imaging

Tomographic MRI using protons does not permit imaging of gases and even of low-density tissue, because the product of polarisation and spin density is too small. X-rays give a better image of the lung, but in both cases it is not possible to study the lung ventilation. As mentioned at the end of Section 4.1, K-edge subtraction with xenon has been tried with moderate success.
At Mainz a new and promising method has been developed in a completely different field: MRI with hyper-polarised $^3$He [26,27]. For ordinary MRI the proton spin polarisation at room (or body) temperature is simply given by the Boltzmann factor and amounts to about $5 \times 10^{-6}$ at 1.5 T. Helium-3 (spin $\frac{1}{2}$) can be polarised to a nuclear spin polarisation of more than 0.5 by optical pumping. The procedure is briefly described: direct optical pumping of the ground state is not possible, because the lowest transition energy ($\sim 20$ eV) would be in the vacuum ultraviolet region. Therefore, the meta-stable $1s2s^3S_1$ state is used. The state is populated in a gas discharge at a pressure of about 1 mbar in a superimposed magnetic field. Then the $F = \frac{3}{2}$ hyperfine level of this state is fully polarised by optical pumping with an infrared laser (1083 nm) resulting in a nuclear polarisation close to one. By excimer formation and decay the nuclear polarisation is transferred to the ground state. The helium is, again in a homogeneous magnetic field, compressed into a specially selected glass cell at three bar. After this manipulation the polarisation is still above 0.5, and the longitudinal relaxation time is about 100 h. The hyper-polarised $^3$He can then be transported to a hospital in a ‘magnetic suitcase’. Slightly modified (to match the lower $^3$He frequencies) standard MRI equipment can then be used for the tomographic imaging. Fig. 16 shows the difference between a non-smokers (left) and a smokers (right) lung ventilation. The bright areas are those with high $^3$He spin density.

The $^3$He comes from tritium decay and is a by-product of Hydrogen bomb production and maintenance. Since the total world production of $^3$He gas is only about 1500 m$^3$, more than 95% of the $^3$He have to be recovered in regular hospital use. As opposed to the Xenon K-edge images the $^3$He images have achieved diagnostic quality.

The example of $^3$He MRI shows also that, if a new project in medical imaging is started, not only the direct competition but also the developments in, at first glance, far lying fields of research should be considered.

7. Detectors for nuclear medicine

Many developments have been going on for detectors in nuclear medicine over the last decade to
improve positron emission tomography (PET) systems in industry and in research institutes, aiming for faster imaging and better resolution [28,29]. While for PET decent $DQE$s can be achieved, this is not the case for single photon emission tomography (SPET), for which the use of collimators limits the $DQE$ to levels of some percent. The only way out here would be a Compton imaging system. There are some occasional and some more permanent efforts in this direction, but, in general, progress in this field seems to be slow and cumbersome [30].

Because of the many nuclides available for SPET and the corresponding large number of biological markers, good SPET detectors are very useful for functional imaging and can be a competition or a complement to morphological imaging.

A general trend is the increased importance of small animal imaging for bio-medical research [31,32]. The new developments for PET and SPET are focused around new scintillators [33,34], like LSO, GSO, YAP and CZT, and new position sensitive photodetectors, like multi-anode photomultipliers, or, just at the beginning, hybrid photodetectors (HPDs) [35,36]. Depth of interaction techniques are being further developed to improve the spatial resolution making use of either spectral coding or decay time coding [37].

8. Detectors for protein crystallography and small angle X-ray scattering

There is no need to discuss the field of detectors for applications in biocrystallography and in small angle X-ray scattering from biological samples here again in depth, because a number of relatively new summary articles have been published to which the reader is referred, e.g. [38–40].

There are presently no detector systems available which make full use of the performance of third generation synchrotron light sources. The available detectors do not permit time resolved studies of dynamic processes or structural analysis of crystals with very large unit cells or large mosaicity. Therefore, in biological structure research there is a need and a challenge for new developments.

9. Summary and outlook

The successful development of imaging detectors requires some understanding of basic imaging theory. Otherwise a reliable comparison of new detectors with existing systems is not possible.

Digital detectors for standard medical X-ray equipment are at the industrial production threshold.

With Si-strip detectors the important step of using them ‘edge-on’ for scanning systems has been made.

The NIKOS system for intravenous coronary angiography with synchrotron radiation is in the medical evaluation phase. About 300 patients have already been examined.

Synchrotron radiation is also used in new promising imaging modalities, namely in phase contrast and in diffraction enhanced imaging, opening up new possibilities in medical diagnostis because of increased object contrast.

Detectors for nuclear medicine have made a big step away from Anger cameras, using new scintillators and position sensitive photodetectors.

In biocrystallography and in small angle X-ray scattering high-performance systems are still urgently required. There are many other aspects in bio-medical research, where new radiation detectors can make important contributions.

New projects, if properly chosen, are always useful and rewarding.

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References

[23] F. Arfelli, private communication.