Two-dimensional X-ray detectors have become a standard at synchrotron sources for practically all applications including tomography. Consequently, a lot of development to optimize these detectors has been, and is still being, done. One can divide area detectors into two main classes: in direct detection the X-ray photons are directly converted to an electrical signal via electron–hole pair creation, either in a semiconductor or in a gas. In indirect detection the X-ray photons are first converted into visible light that is subsequently converted to an electronic signal in the semiconductor.

Direct detection has certain advantages. It avoids the conversion step into visible light that can lead to degradations of the recorded image, as will be explained in detail in this chapter. It also allows for single-photon counting, where each individual X-ray photon is detected and processed, resulting in a virtually noise-free detector that is sensitive to the energy of the incident X-rays. Such detectors are under development and require quite sophisticated electronics.

A disadvantage of direct detection is that commercial sensors such as charge-coupled devices (CCD), being mainly developed for visible-light detection, cannot be used because of their low sensitivity in the X-ray regime. Other disadvantages of photon counting are the need for sophisticated electronics and the fact that time is needed to process each photon, which limits the incident flux that can be detected. The alternative is to convert the X-ray photons to visible light and to use integrating detectors such as CCDs. So far, the most effective, low-cost and low-risk solution for tomography has been indirect detection, where a scintillator converter screen is optically coupled to an integrating detector, either a commercial CCD camera, or a specially developed amorphous photodiode array. In this chapter we will explain this technology and discuss the various components. We will start with a survey of the various existing types of X-ray conversion screens. This is followed by a discussion of the optical relays and the light sensors required to convert the light into a digital image.

A large number of beamlines at synchrotron radiation sources use tomography X-ray imaging techniques in an increasingly wide range of research, including materials science, medical, geophysical, environmental or palaeontological studies. Consequently, both the input field of view (FOV) and the required spatial resolution span a wide range. For instance, the largest FOVs for tomography at synchrotron radiation sources is at medical beamlines and can be as large as 300 mm × 20 mm. The currently smallest FOV is 300 × 300 µm² with
Fig. 10.1. The electronic band structure of an activated inorganic crystal scintillator. When an X-ray photon interacts with the scintillator material, electrons move from the valence band to the conduction band. It can fall down into one of the excited states of the activator without emission. When it is transmitted from the activator energy band back to the ground state, a photon is emitted. These de-excitations are in the visible-light band.

$2048 \times 2048$ pixels each of $0.14 \mu m$ (Koch et al., 1999a). This large range of three orders of magnitude necessarily implies the use of different technologies and materials but they are almost all based on the principle of an X-ray converter screen optically relayed by either lens coupling or tapered fibre-optic coupling to commercial CCD cameras. The exception is the large panel imagers developed for the medical imaging market, which will be briefly discussed at the end of the chapter.

10.1 Scintillation mechanism

Detectors used for tomography at synchrotron beamlines use a converter screen to convert X-ray photons to visible-light photons that are subsequently detected by a photodetector, mostly of the CCD type. This conversion is done via a scintillation or luminescence process, and therefore these converter screens are often called scintillation screens. The converter screens used are made of inorganic scintillators that have a high density and a high effective atomic number, resulting in high stopping power and high light yield for X-rays photons. In the so-called intrinsic scintillators the luminescence is produced by part of the crystal lattice itself. These scintillation materials have an energy-band diagram similar to that of semiconductors, with a valence band separated by an energy gap from the conduction band. When an X-ray photon is absorbed energy is deposited in the crystal and electrons in the valence band can be excited to the conduction band, leaving holes in the valence band. When an excited electron falls back to the valence band and recombines with a hole the excess energy is dissipated by
emission of photons with an energy equal to the width of the bandgap. For intrinsic scintillators the efficiency of the de-excitation process is, however, low and the bandgaps are, in general, too large to emit light in the visible range that is needed for detection by common CCDs. To overcome this limitation scintillators are doped with typical fluorescent ions such as europium (Eu), terbium (Tb) or cerium (Ce). Inorganic scintillators doped with rare-earth elements are thus suitable because they exhibit a relatively large bandgap, good light yield, fast decay time and a minimal self-absorption. Doping with these activators makes the scintillation process more efficient and creates energy levels inside the energy gap producing scintillation light in the visible region (see Fig. 10.1). These doped scintillators are the ones used to construct the 2D detectors for tomography at synchrotron beamlines.

10.2 Spatial resolution and detective quantum efficiency

Two important characteristics of an imaging system, in particular for tomography, are the spatial resolution and the detective quantum efficiency (DQE). We will first discuss the various parameters that determine the spatial resolution, secondly we will analyse the contributions of the various components to the DQE. It will be shown that these are often two competing requirements.

A schematic layout of the most commonly used detector principle for tomography at synchrotron beamlines is given in Fig. 10.2. In this case, the converter screen (or scintillator) is mounted on a substrate, which will be explained in more detail in Section 10.4. In such a system the spatial resolution is determined by the X-ray interaction in the scintillator, by the X-ray interaction in the substrate, and by the performance of the optics. We will first discuss the X-ray interaction in the scintillator and the substrate.
As pointed out in Chapter 4 there are various kinds of X-ray interactions with solids, namely the photoelectric effect, (elastic) Rayleigh scattering and (inelastic) Compton scattering. For high energies above 150 keV and a yttrium aluminium garnet (YAG) crystal the Compton effect becomes dominant. The scattered photon, the energy and direction of which are governed by the Compton scattering distribution, can escape, be scattered again or be absorbed. The resulting electron range (the distance at which half of the particles have been absorbed) which is very short for low-energy photoelectrons becomes an appreciable fraction of the scintillator thickness at higher Compton electron energies. The assumption of total electron energy absorption at the interaction site is less accurate. An increasing part of the electron path takes place outside the interaction volume, resulting in the loss of energy absorbed in the ‘phosphor’ and/or a decrease in the spatial resolution. As an example, the average path length travelled by an electron in a YAG crystal is $\approx 2.8 \mu m$ at 20 keV and $\approx 44 \mu m$ at 100 keV (National Institute of Standards and Technology, 2006). This shows that the effect can be an important source of degradation whenever the Compton effect is dominant. Compton scattering is negligible as a source of background noise, however, in imaging applications for X-ray energies below 100 keV (Hoheisel et al., 2004).

The predominant physical process that reduces the spatial resolution at low X-ray energies is the fluorescence generated by photoelectric absorption. Monte Carlo simulations of photon and electron transport have been made for a YAG crystal (Koch et al., 1998). The Integral Tiger Series (ITS) code uses a Monte Carlo simulation of photon and electron transport in matter to derive the radial absorbed dose distributions. The calculation used a narrow and parallel pencil beam of 14-keV X-rays on a 5-µm thick YAG:Ce scintillator supported by a 100-µm thick undoped YAG substrate. The simulation, which did not take into account the scattering of the emitted visible light, shows a very high absorbed dose in 100 nm full width at half-maximum (FWHM) due to the short path of the Auger electrons. The energy deposited outside these 100 nm is due to secondary electrons and characteristic fluorescence X-rays. On the other hand, calculations at 30 and 100 keV have shown a point-spread function (PSF) tail of the radial energy distribution that degrades the spatial resolution. The small FWHM of the PSF at low energies indicates that the spatial resolution is in this case not limited by the scintillator but by the resolving power of the optics, i.e. in particular of the microscope objective, which will be discussed later.

Another source of tails in the PSF is the undesired luminescence from the substrate, e.g. by Ce impurities in undoped YAG. Light emission from substrates has been measured (Martin et al., 2005) and corresponds in the worst case to 20% of the luminescence of the YAG:Ce epitaxial layer. This becomes even more important as thinner layers of scintillator are used. With the following light yields\(^6\)

\[^6\]Number of emitted visible photons emitted in 4$\pi$ steradian per unit of absorbed X-ray energy.
measured by Martin et al. (2005) – 11 ph/keV for LAG:Eu and 1.2 ph/keV for undoped YAG (which corresponds to 6% of the typical value of doped YAG:Ce) – a 170-µm undoped YAG substrate emits more light than a 5-µm LAG:Eu layer for X-ray energies above 20 keV. This means that at higher energies an undoped YAG substrate does not allow the use of very thin LAG:Eu layers. Therefore, LAG and YAG scintillators layers grown by liquid phase epitaxy on a YAG substrate are not ideal for high spatial resolution at high energies. In the case of LAG:Eu layers on a YAG substrate one could use colour filters to suppress about half of the substrate’s luminescence signal. This solution is not possible for the YAG:Ce layer because the YAG substrate luminescence wavelength is too close to the YAG:Ce layer emission line. Other solutions are to use purer YAG substrates or some other deposition techniques such as the sol-gel method, which permits a larger choice of substrate materials.

Another component that determines spatial resolution is the optical coupling between the converter screen and the CCD. The CCD camera is focused onto the object plane inside the scintillator located at the distance \( z_0 \) from the surface (see Fig. 10.2). All other planes in front of and behind the object plane are out of the focus but nevertheless contribute to the total light projected onto the CCD. The image resolution is determined by the defect of focus \( \delta z \) of the image distribution outside the object plane. Other degradations of the image are due to diffraction and spherical aberrations arising from the thickness of the scintillator \( z \) and the substrate \( t \). The effects affecting spatial resolution are the following:

- defect of focus: \( R \approx \delta z \cdot \text{NA} \),
- diffraction: \( R \approx \lambda/\text{NA} \),
- spherical aberration: \( R \approx t \cdot \text{NA}^3 \),

where \( \delta z \) is the defect of focus, \( \lambda \) is the wavelength of light and \( \text{NA} \) is the numerical aperture. The spatial resolution \( R \) determined by the first two effects can be calculated by numerical simulations (Koch et al., 1998) and can be fitted by:

\[
R_{\text{FW50\%int}} = \sqrt{(p/\text{NA})^2 + (q \cdot z \cdot \text{NA})^2},
\]

where \( R \) and \( z \) are in micrometres. The coefficients \( p = 0.18 \) and \( q = 0.075 \) (Koch et al., 1998) are calculated for a 50% integrated line-spread function (LSF). Here, the first term is due to diffraction and the second to the defect of focus. The spatial resolution is given in Fig. 10.3 as a function of numerical aperture for different screen thicknesses.

The other parameter, besides spatial resolution, which is important for imaging systems is the detective quantum efficiency, DQE. This parameter relates the output to the input signal-to-noise ratio:

\[
\text{DQE} \equiv \frac{\text{SNR}_\text{out}^2}{\text{SNR}_\text{in}^2} \approx \eta_{\text{abs}} \cdot \left( 1 + \frac{1}{\eta_{\text{abs}} \cdot \left( \frac{\eta_{\text{x/e}}}{\eta_{\text{coll}}} \cdot \frac{1}{\frac{E_X}{E_e}} \right) \cdot \eta_x/\nu} \right)^{-1},
\]

(10.1)
Detectors for synchrotron tomography

Fig. 10.3. Spatial resolution versus numerical aperture (NA) of an optical system for different thicknesses of the single-crystal scintillator. The parameters used for the YAG:Ce scintillator are: refractive index \( n = 1.95 \) and X-ray wavelength \( \lambda = 550 \text{ nm} \). One way to increase the spatial resolution is to reduce the thickness, which in turn reduces the absorption efficiency.

Table 10.1. Overview of commercially available CCD cameras.

<table>
<thead>
<tr>
<th>CCD manufacturer</th>
<th>pixel size</th>
<th>array size</th>
<th>URL</th>
</tr>
</thead>
<tbody>
<tr>
<td>Apogee</td>
<td>9 ( \mu \text{m} )</td>
<td>1536 \times 1024</td>
<td><a href="http://www.ccd.com">www.ccd.com</a></td>
</tr>
<tr>
<td>Frelon</td>
<td>14 ( \mu \text{m} )</td>
<td>2048 \times 2048</td>
<td><a href="http://www.esrf.fr">www.esrf.fr</a></td>
</tr>
<tr>
<td>PCO</td>
<td>7.4 ( \mu \text{m} )</td>
<td>2048 \times 2048</td>
<td><a href="http://www.pco.de">www.pco.de</a></td>
</tr>
<tr>
<td>Roper</td>
<td>7.4 ( \mu \text{m} )</td>
<td>1600 \times 1200</td>
<td></td>
</tr>
<tr>
<td></td>
<td>6.8 ( \mu \text{m} )</td>
<td>1317 \times 1035</td>
<td><a href="http://www.roperscientific.com">www.roperscientific.com</a></td>
</tr>
<tr>
<td></td>
<td>6.45 ( \mu \text{m} )</td>
<td>1392 \times 1040</td>
<td></td>
</tr>
<tr>
<td>Photonic</td>
<td>6.45 ( \mu \text{m} )</td>
<td>1392 \times 1040</td>
<td><a href="http://www.photonic-science.co.uk">www.photonic-science.co.uk</a></td>
</tr>
<tr>
<td>ATMEL</td>
<td>14 ( \mu \text{m} )</td>
<td>2048 \times 2048</td>
<td><a href="http://www.atmel.com">www.atmel.com</a></td>
</tr>
<tr>
<td>Pixel Vision</td>
<td>14 ( \mu \text{m} )</td>
<td>2048 \times 2048</td>
<td><a href="http://www.pvinc.com">www.pvinc.com</a></td>
</tr>
<tr>
<td>Dalsa</td>
<td>12 ( \mu \text{m} )</td>
<td>1024 \times 1024</td>
<td><a href="http://www.dalsa.com">www.dalsa.com</a></td>
</tr>
<tr>
<td></td>
<td>14 ( \mu \text{m} )</td>
<td>1024 \times 1024</td>
<td></td>
</tr>
</tbody>
</table>

where \( \text{SNR}_{\text{out}} \) and \( \text{SNR}_{\text{in}} \) are the input and output signal-to-noise ratios of the detector, \( \eta_{\text{abs}} \) is the absorption efficiency for X-rays in the scintillator, \( \eta_{\nu/e} \) is the quantum efficiency of the CCD, \( \eta_{\text{coll}} \) is the collection efficiency of light by an objective with numerical aperture NA in combination with a scintillator with a refractive index \( n \), \( E_X \) is the X-ray energy, \( E_{\nu} \) is the photon energy of the visible-light photons and \( \eta_{X/\nu} \) is the conversion efficiency of X-rays to visible-
light photons (light yield), meaning the fraction of the incident X-ray energy that appears as scintillation light. The DQE, ranging between 0 and 1, is reduced if the imaging system degrades the spatial resolution, if the detector adds noise to the final image, or if not all incoming X-rays are absorbed. In order to improve the absorption of thin scintillators they are often used with X-rays above the K-absorption edge (17.04 keV for YAG:Ce, 50.24 keV for GGG and 63.3 keV for LAG:Eu). While this improves the absorption efficiency, it also increases X-ray fluorescence, which leads to a loss of spatial resolution as explained above. For systems with sufficiently high NA the DQE is mainly determined by the absorption efficiency. As shown in Tab. 10.2, the quantum efficiency of the CCD camera and the type of scintillator have an impact on the DQE at low NA. For example, GGG:Eu and LAG:Eu layers have the same absorption efficiency but different light yields, and we note a DQE improvement for NA=0.3 of 20% and 23% with the FReLoN and Dalsa, respectively. Table 10.1 lists these two and other CCD cameras and specifies their resolution. At higher numerical aperture, the improvement is less significant: 8% and 12.5% for FReLoN and Dalsa cameras, respectively. A promising new camera for fast microtomography is the new fast Dalsa 1M60P that uses a FTT1010M CCD chip and is more sensitive than the 1M60 type. With a LAG:Eu converter layer and a NA=0.3 it has a DQE of 0.027, which is an improvement of 58% over the DALSA 1M60. As stated before, at high numerical apertures the DQE is mainly determined by the absorption efficiency of the converter screen and less by the choice of the CCD camera. For example, although the KAF-4320E Kodak CCD chip has a high quantum efficiency (70% at 580 nm) that is 150% more sensitive than the ATMEL (see Tab. 10.1) CCD chip, the improvement of the DQE at NA=0.7 is weak, 0.09 instead of 0.076 with LAG:Eu layer.

10.3 Powder screens

In the previous section we analysed the characteristics of the detector that determine the overall performance, in particular the spatial resolution and the detective quantum efficiency. It was shown that the converter or scintillator screen is of crucial importance for high-resolution tomography. Converter screens can be classified into two categories: powder converter screens and crystal converter screens. In this section we will discuss powder screens, in the next crystal screens.

In powder converter screens a small-grained 'phosphor' powder is mixed with a binding agent and is processed to a film. The two main powder screens used for X-ray detection at synchrotrons are Gd$_2$O$_2$S:Tb, called P43 or Gadox and Y$_3$Al$_5$O$_{12}$:Ce called P46. Although the packing density of the powders is half that of the solid material Gadox is, nevertheless, very attractive for its high density and its X-ray to visible-light yield ($\approx$60 photons/keV). A disadvantage of P43 (Gadox) is its long light decay time of a few hundred microseconds, which severely limits its application in fast imaging applications. In this case, P46 is preferred for its short decay time. Screens based on powder phosphors are commercially available down to a grain size of approximately 1 µm. Their resolution in terms
Table 10.2. Characteristics of some scintillator materials. The absorption efficiency and DQE are calculated at 20 keV with a 5-μm thick scintillator layer using eqn 10.1. The resolution is deduced from experiment. Cloetens et al. (1999) has measured a width of the LSF of 2 μm with NA=0.3 and 25 μm YAG:Ce. A spatial resolution of 0.8 μm has been measured by Wang et al. (2001) with a Nikon plane achromat objective with 40× magnification, NA=0.6 and 5 μm YAG:Ce layer. Stampanoni et al. (2002) have measured a spatial resolution of 1.04 μm with a 1.8-μm thick YAG:Ce scintillator and 20×, NA=0.7 Olympus optic.

<table>
<thead>
<tr>
<th>optic</th>
<th>scintillator</th>
<th>absorption</th>
<th>DQE FReLoN</th>
<th>DQE Dalsa</th>
<th>spatial resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>10×</td>
<td>YAG:Ce</td>
<td>0.07</td>
<td>0.021</td>
<td>0.013</td>
<td>2 μm (1)</td>
</tr>
<tr>
<td>NA=0.3</td>
<td>LAG:Eu</td>
<td>0.11</td>
<td>0.031</td>
<td>0.017</td>
<td></td>
</tr>
<tr>
<td></td>
<td>GGG:Eu</td>
<td>0.11</td>
<td>0.037</td>
<td>0.021</td>
<td></td>
</tr>
<tr>
<td>20×</td>
<td>YAG:Ce</td>
<td>0.07</td>
<td>0.049</td>
<td>0.038</td>
<td>1.04 μm (2)</td>
</tr>
<tr>
<td>NA=0.7</td>
<td>LAG:Eu</td>
<td>0.11</td>
<td>0.076</td>
<td>0.056</td>
<td></td>
</tr>
<tr>
<td></td>
<td>GGG:Eu</td>
<td>0.11</td>
<td>0.082</td>
<td>0.063</td>
<td></td>
</tr>
</tbody>
</table>

(1) Cloetens et al. 1999
(2) Stampanoni et al. 2002

of FWHM of the line-spread function is, however, approximately equal to their thickness (Swank, 1973).

Homogeneous screens of 2 to 3 μm thickness and similar resolution can be deposited. Smaller grain sizes are manufactured with 100 nm diameter but these screens show powder agglomerates as well as a poor X-ray to light conversion yield (Koch et al., 1999b).

10.4 Crystal converter screens

In the previous section it was mentioned that the spatial resolution of powder converter screens is equal to the thickness of the screen, giving an unfavourable compromise between X-ray stopping power and spatial resolution. To overcome this limitation, thin single-crystal scintillators and single-crystal films (SCF) have been developed and are now commercially available in a thickness ranging from 1 μm to 50 μm. Due to the better lateral spatial resolution compared to powder screens, thin-film scintillators have become the main converter screen technology for high-resolution X-ray imaging. There are many methods for deposition of luminescent films, including pulsed laser deposition, sputtering, liquid phase epitaxy and the sol-gel process. The latter technique is suitable for the deposition of amorphous and crystalline thin films through dip- and spin-coating. However, a detailed treatment of the various thin-film growth techniques is outside the scope of this book. In Section 10.4.1 we will discuss the essential properties of crystal converter screens. In Section 10.4.2 the most commonly used materials will be discussed.
10.4.1 Essential properties of crystal converter screens

A number of properties of crystal converter screens are of essential importance for high-resolution X-ray imaging, namely X-ray absorption, visible-light emission, optical and engineering properties. We will discuss them separately.

- X-ray absorption can be maximized by a combination of high-density material (>5 g/cm$^3$) with a large atomic number (>50). The X-ray absorption efficiency in a scintillator is proportional to $\rho Z^n$ with $\rho$ the density, $Z$ the effective atomic number and $n$ varying between 4 (low energies <500 keV) and 5 (high energy), see Section 4.2.3.

- The visible-light emission has various aspects. First, one wants a high light yield (>15 visible photons/keV). The emission efficiency is defined as the ratio of the output light energy to absorbed radiation energy. Secondly, the emission wavelength has to be well matched to the peak sensitivity of the CCD chip used, which is generally at wavelengths between 550 and 650 nm. In most cases the emission peak wavelength is determined by the activator material and one thus has some flexibility by using different doping materials. Thirdly, one desires rapid decay and short afterglow (2 ms) of the crystal converter screen. Decay and afterglow produce the same phenomenon – image lag – but have a different origin. Decay reflects the time that it takes for the excited energy states to be de-excited. It is determined by the scintillation material and the type of activator. Afterglow is due to another mechanism. When electrons are excited by radiation, some excited electrons are activated to an energy level from which the direct transition to the valence band is forbidden. These electrons have to be further excited to a higher energy state before they can return to the valence band. Such a mechanism can cause severe lag in the light emission. Afterglow can be caused by the material, by imperfect processing, unintentional doping, etc. Both decay and afterglow are harmful to synchrotron tomography applications in cases where high frame rates are essential and scintillator afterglow limits the useful dynamic range of the detector. Furthermore, the light output should be linear with the incident X-ray flux.

- The optical properties required are a high transmittance of the emitted visible light with no scattering.

- Finally, the crystal converter screen should have a good mechanical strength, preferably be non-toxic, be easy to machine and should be radiation hard, meaning no degradation caused by X-ray irradiation.

An overview of the key parameters for different crystal converter screens is given in Tab. 10.3.

10.4.2 Bulk converter screens

Crystal converter screens are either used in bulk crystalline form or in thin-film-on-substrate form. We will first describe the most important bulk crystals used, followed by the most suitable thin-film converter screens.
Table 10.3. Characteristic data of scintillators used for X-ray imaging.

<table>
<thead>
<tr>
<th>Material</th>
<th>Name</th>
<th>Density (g/cm³)</th>
<th>Zeff</th>
<th>λ (nm)</th>
<th>LY (ph/keV)</th>
<th>Refractive Index</th>
<th>Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gd₂O₂S:Tb</td>
<td>P43</td>
<td>7.3</td>
<td>59.5</td>
<td>545</td>
<td>60</td>
<td>1.8</td>
<td>powder</td>
</tr>
<tr>
<td>Y₃Al₅O₁₂:Ce</td>
<td>P46</td>
<td>4.55</td>
<td>32</td>
<td>530</td>
<td></td>
<td></td>
<td>powder</td>
</tr>
<tr>
<td>CsI(Tl)</td>
<td></td>
<td>4.53</td>
<td>54</td>
<td>550</td>
<td>65</td>
<td>1.8</td>
<td>crystal</td>
</tr>
<tr>
<td>Bi₄Ge₃O₁₂</td>
<td>BGO</td>
<td>7.13</td>
<td>75</td>
<td>480</td>
<td>8.2</td>
<td>2.15</td>
<td>crystal</td>
</tr>
<tr>
<td>CdWO₄</td>
<td></td>
<td>7.90</td>
<td>64</td>
<td>530</td>
<td>15</td>
<td>2.25</td>
<td>crystal</td>
</tr>
<tr>
<td>Lu₃Al₅O₁₂</td>
<td>LAG:Ce</td>
<td>6.73</td>
<td>63</td>
<td>550</td>
<td></td>
<td></td>
<td>crystal</td>
</tr>
<tr>
<td>Y₃Al₅O₁₂:Ce</td>
<td>YAG:Ce</td>
<td>4.55</td>
<td>32</td>
<td>550</td>
<td>40</td>
<td>1.82</td>
<td>crystal</td>
</tr>
<tr>
<td>Lu₂SiO₅:Ce</td>
<td>LSO:Ce</td>
<td>7.4</td>
<td>66</td>
<td>420</td>
<td>25</td>
<td>1.82</td>
<td>crystal</td>
</tr>
<tr>
<td>Y₃Al₅O₁₂:Ce</td>
<td>YAG:Ce</td>
<td>4.55</td>
<td>32</td>
<td>550</td>
<td>20</td>
<td>1.82</td>
<td>SCF</td>
</tr>
<tr>
<td>Lu₃Al₅O₁₂:Eu</td>
<td>LAG:Eu</td>
<td>6.73</td>
<td>63</td>
<td>595</td>
<td>11</td>
<td></td>
<td>SCF</td>
</tr>
<tr>
<td>Gd₃Ga₅O₁₂:Eu</td>
<td>GGG:Eu</td>
<td>7.1</td>
<td>53</td>
<td>595</td>
<td>44</td>
<td>1.96</td>
<td>SCF</td>
</tr>
<tr>
<td>Lu₂O₃:Eu⁺⁺</td>
<td></td>
<td>8.4</td>
<td>68.8</td>
<td>611</td>
<td>20</td>
<td>1.88</td>
<td>polycrystal</td>
</tr>
<tr>
<td>Gd₂O₃:Eu³⁺</td>
<td></td>
<td>7.1</td>
<td>61</td>
<td>611</td>
<td>19</td>
<td>1.82</td>
<td>polycrystal</td>
</tr>
</tbody>
</table>

Application of crystal converter screens in X-ray medical imaging has resulted in rapid development of high-density bulk single-crystal scintillators. Polishing and classical cut technologies are used to prepare single-crystal scintillators for X-ray tomography applications.

10.4.2.1 Bismuth germanate One system is bismuth germanate (Bi₄Ge₃O₁₂, or commonly abbreviated as BGO). BGO has a high density and a large atomic number. Unfortunately, the poor light yield, difficulties in obtaining a smooth surface as well as its high refractive index are disadvantages for high-resolution imaging (Tseng et al., 2000). Therefore, BGO seems to be more suited for high gamma energy counting probes than for X-ray tomography.

10.4.2.2 Cadmium tungstate Another system is cadmium tungstate (CdWO₄, sometimes abbreviated as CWO) which is a suitable material for various applications due to its high X-ray stopping power, its high light yield with little afterglow, emission near 470 nm well matched to the sensitivity of CCD detectors and the fact that it is non-hygrosopic. However, machining and thinning of CdWO₄ single-crystals is not easy due to its intrinsic (010) cleavage plane. As a result it is not possible to find commercial CdWO₄ scintillators much thinner than 50 µm. Lee et al. (1997) have used a 60-µm thick CdWO₄ single crystal and optics with a numerical aperture of 0.5. The spatial resolution obtained was about 1.5 µm at 20 keV. Jung et al. (2002) have used a CdWO₄ single-crystal scintillator cleaved to a thickness less than 100 µm in a pink beam (non-monochromatic X-ray beam with 6 to 30 keV energy). The spatial resolution was 2.9 µm.

To obtain a better spatial resolution, the scintillator must be thinner. A sol-gel processing technique has been applied to produce a thin film of cadmium tungstate (Lennstrom et al., 2003). The resulting film consists of crystal grains of approximately 1 µm in diameter, which is a limitation for high-resolution imaging.
10.4.2.3 **Lutetium oxyorthosilicate**  A promising bulk-crystal scintillator is cerium-doped, high-density, high-Z lutetium oxyorthosilicate (LSO). Although the absorption efficiency of this scintillator is lower than that of BGO it appears to have a better light yield than BGO and a shorter decay time than Eu-doped scintillators. LSO is available from CTI/USA\(^7\). Hamamatsu\(^8\) used this scintillator in a high-resolution X-ray imaging camera installed at SPring8. The single-crystal scintillator is 10\(\mu\)m thick and is glued to an amorphous carbon plate. Uesugi \textit{et al.} (2001) have achieved 1\(\mu\)m spatial resolution. Spatial resolution is limited by the 10\(\mu\)m thickness of the crystal, which cannot be reduced any further by mechanical polishing.

10.4.3 **Composed converter screens**

In order to get even thinner converter screens one has to deposit a single-crystal or a polycrystalline film on a substrate. There are various different methods to grow thin films on a substrate, and the growth mechanism can have a substantial effect on the performance of the thin-film converter screens. However, this subject is outside the scope of this book and we will only treat the physical properties of the most commonly used films.

10.4.3.1 **YAG:Ce on YAG**  Yttrium aluminium garnet (YAG:Ce) has promising scintillation properties when doped with cerium. The spectrum is unusual for a cerium-doped scintillator and the peak in the emission spectrum occurs at 550 nm. This longer wavelength is a good match to the quantum efficiency of CCD cameras.

10.4.3.2 **LAG:Eu on YAG**  Lutetium aluminium garnet (LAG:Eu) has an absorption efficiency 8 times higher at 15\(\text{keV}\) and 2 times higher at 40\(\text{keV}\) than YAG:Ce. However, the growth process for LAG on undoped YAG needs to accommodate the lattice mismatch between LAG and YAG and can be achieved by partial substitution of aluminium by scandium in LAG. Unfortunately, LAG doped by Eu and Sc has a considerably lower light yield compared to an epitaxial layer of YAG:Ce. The principal cause of this smaller light yield is the loss of excitation energy on exciting the luminescent centers formed by Sc (Martin \textit{et al.}, 2005). Even though YAG:Ce has a faster decay than LAG:Eu, which is of importance for fast imaging applications, LAG:Eu films are preferred since they have a higher X-ray stopping power and a lower light output, resulting in a net gain of a factor 4 for a given exposure time above the absorption edge of lutetium (63\(\text{keV}\)). Opportunities for improving X-ray detectors in terms of spatial resolution are somewhat limited as they are already near the theoretical limits of 0.4\(\mu\)m. The best performance measured is 0.5\(\mu\)m FWHM obtained with 1-\(\mu\)m LAG:Eu (Lu\(_3\)Al\(_5\)O\(_{12}\):Eu) thick film and NA=0.95 (Koch \textit{et al.}, 1999\textit{a}).

\(^7\text{www.siemens.com.}\)

\(^8\text{www.hamamatsu.com.}\)
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Fig. 10.4. Absorption efficiency of more current scintillators used for X-ray imaging on synchrotron radiation sources for 5-µm thick layers: Lu₃Al₅O₁₂ (LAG), Gd₃Ga₅O₁₂ (GGG), Lu₂SiO₅ (LSO), Y₃Al₅O₁₂ (YAG) and Gd₂O₂S:Tb (P43).

Possible improvements are in X-ray stopping power, light yield (and thus DQE) and decay time.

10.4.3.3 GGG:Eu Gadolinium gallium garnet (GGG:Eu) shows excellent scintillator properties compared to YAG:Ce and LAG:Eu. Its absorption efficiency is comparable to LAG:Eu material (see Fig. 10.4) but its light yield is better than YAG:Ce and more than 2 times better than LAG:Eu. In addition, the afterglow is fast compared to LAG:Eu and allows resolution of a dynamic range between 16 and 17 bit in a successive image mode (Martin et al., 2005) which is a significant improvement over the 10-bit dynamic range obtained with a YAG:Ce layer (Koch et al., 1999b).

10.4.4 Polycrystalline scintillators

Both lutetium oxide (Lu₂O₃:Eu³⁺) and gadolinium oxide (Gd₂O₃:Eu³⁺) doped with europium represent a dense version of Y₂O₃:Eu³⁺ phosphor. Lu₂O₃:Eu³⁺ has an absorption efficiency that is ≈2 times higher at 15 keV, a light yield ≈2 times higher than LAG:Eu and its emission line shows a strong peak at 610 nm, which is well adapted to a CCD chip. Lu₂O₃:Eu³⁺ and Gd₂O₃:Eu³⁺ have been prepared by sol-gel processing. This technique is suitable for the deposition of amorphous and crystalline thin films through dip- or spin-coating. Researchers at
the University of Lyons (Garcia-Murillo et al., 2003) have studied dip-coating for X-ray imaging applications. Multicoating (up to 50 coatings) and heat treatments are required to obtain 800-nm thick films. The layers are finally annealed for 1 h at 1000 °C for high densification and crystallization of the materials. A definite advantage of the sol-gel method as compared to liquid phase epitaxy is a larger choice of substrate material for crystal growth. Deposition on silicon and silica has been realized by Garcia-Murillo et al. (2002). However, this technique is limited in terms of thickness (1 µm) due to the mechanical stresses between layers. Dujardin et al. (2005) have also deposited the same material by pulsed laser deposition (PLD) and they have shown the possibility of depositing a thicker layer by PLD, but further development is necessary to obtain a homogeneous layer.

All the described screens have been used in tomographic applications at synchrotron beamlines. The commercially available inorganic scintillators, YAG:Ce, CdWO₄ and LAG:Eu were successfully used by Stampanoni et al. (2002), Carlo et al. (2001), Koch et al. (1998) and Beckmann (2001). However, for imaging experiments with higher X-ray energies LSO:Ce was preferred by Yagi et al. (2004a). Finally, GGG:Eu seems to be preferable over YAG:Ce and LAG:Eu for various reasons: the light yield of LAG:Eu is low and the undoped YAG substrates for the YAG:Ce and LAG:Eu layers emit undesired luminescence that degrades the spatial resolution (Martin et al., 2005).

### 10.5 Optical coupling

In Sections 10.3 and 10.4 we discussed the screens that convert the incident X-rays to visible-light photons. In this section, we will discuss the optical coupling between these converter screens and the photodetector. Two ways to achieve this optical coupling will be discussed: lens coupling and fibre-optic (FO) coupling. For the latter, a taper contains millions of individual fibres (core + cladding) that transmit an element of the image created on the powder screen. Fibres are clad to guide the light by internal reflection. We will start with a comparison of the efficiency between lens and FO coupling for various magnifications and demagnifications. After that, a number of important issues related to lens coupling are treated.

Lens coupling is the simplest and most convenient way and works well for systems with a high spatial resolution. For powder converter screens (Section 10.3) the emitted light is that of a so-called Lambertian source, see Fig. 10.11(a), and the transmission efficiency through the lens coupling is given by (Liu et al., 1994):

\[ \eta = \frac{T_L M^2}{M^2 + 4f^2(1 + M)^2}, \]

where \( f \) is the ratio of the focal length to the effective lens diameter, \( T_L \) is the transmission factor of the lens and \( M \) is the magnification ratio, i.e.
Fig. 10.5. Effect of magnification factor on the collection efficiency for lens coupling.

\[ M = \frac{\text{image size}}{\text{object size}}. \]

For a crystal converter screen (Section 10.4) the source is non-Lambertian, see Fig. 10.11(b), and the transmission efficiency is:

\[ \eta = \frac{T_L M^2}{16n^2f^2(1 + M)^2}, \]

where \( n \) is the optical refractive index of the crystal converter screen. Collection efficiencies (Liu et al., 1994) of lens-coupled systems are plotted in Fig. 10.5. FO coupling on the other hand is preferred when large fields of view are needed. In this case, powder converter screens are mostly used and the transmission of a the tapered fibre optic can be expressed by (Hejazi and Trauernicht, 1997):

\[ \eta = \left( \frac{1}{m} \right)^2 \left( \frac{(n_2^2 - n_3^2)^{1/2}}{n_1} \right)^2 T_F(1 - L_R)F_c, \]

where \( m \) is the demagnification factor, \( n_2 \) is the core index, \( n_3 \) is the cladding index, \( n_1 \) is the external index, \( T_F \) is the transmission of the fibre core, \( L_R \) is the loss at the surface due to Fresnel reflection and \( F_c \) is the fill factor of the fibre core.

In Tab. 10.4 the collection efficiency for both lens coupling and fibre-optic coupling are shown for comparison. Lens coupling is more efficient than FO coupling when the object is magnified because in that case the solid angle is large enough to collect the light coming from the luminescent screen. This is the
Table 10.4. Efficiency of f/1.2 lens and fibre-optic coupling at various magnifications calculated for a Lambertian source, i.e. for a powder screen. Calculated with $T_L = 1$ and $T_F = 1$.

<table>
<thead>
<tr>
<th>magnification</th>
<th>lens coupling efficiency</th>
<th>fibre-optic efficiency</th>
</tr>
</thead>
<tbody>
<tr>
<td>20×</td>
<td>18%</td>
<td>-</td>
</tr>
<tr>
<td>10×</td>
<td>17%</td>
<td>-</td>
</tr>
<tr>
<td>4×</td>
<td>13.7%</td>
<td>-</td>
</tr>
<tr>
<td>1×</td>
<td>4.1%</td>
<td>84%</td>
</tr>
<tr>
<td>0.5×</td>
<td>1.9%</td>
<td>21%</td>
</tr>
<tr>
<td>0.25×</td>
<td>0.7%</td>
<td>6%</td>
</tr>
</tbody>
</table>

basis of normal microscopy and it is also used for synchrotron microtomography. The situation is different for demagnification, where the efficiency of lens coupling drops with increasing demagnification. This is because luminescent screens give off light distributed over an angle of 180° (half-space, Lambert distribution), and lenses transmit only a relatively small fraction of the emitted light. In this case, fibre-optic coupling is more efficient. As an example, a non-magnifying (1:1) optic of the relative aperture $f/1.2$ has an efficiency of 4.1%, while a high-quality non-magnifying FO shows an efficiency of up to 84%. Fibre-optic tapers can thus boost the light collected by a CCD by 20 times compared to a f/1.2 lens and by 14 times compared to a f/1.0 lens.

10.5.1 Lens coupling: finite-focused versus infinity-focused systems

Lens coupling is the preferred optical relay for the magnifying used in microtomography. For high-resolution detectors based on microscope objectives one can adopt two strategies depending on whether one uses finite-focused or infinity-focused optical systems. A finite optical system, or fixed tube length, is an optical system in which the image is formed by the objective alone; the distance between the microscope objective shoulder and the image seat in the phototube is fixed. This distance is named the tube length of the microscope and the tube length is standardized to 160 mm. The situation is different for the infinity-corrected optical systems, where the image is formed by a combination of objective and tube lenses with the microscope objective producing a flux of parallel visible light, imaged at infinity. The tube lens forms an intermediate image in the tube. The magnification produced by an infinity-corrected system is calculated by the ratio of objective focal length and tube lens focal length. Finite optical microscopes are less and less commercially available. Spindler and Hoyer\textsuperscript{9} sells refractive objectives and Ealing\textsuperscript{10} offers reflecting objectives, both with a finite tube length of 160 mm. The situation is different for infinity-corrected optical systems that are sold by different companies that do not use the same tube.

\textsuperscript{9}Linos, Koenigsalle 23, D-37081 Goettingen; www.linos.com.

\textsuperscript{10}Ealing, 3845 Atherton Road, Suite #1, Rocklin, CA 95765; www.ealingcatalog.com.
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Fig. 10.6. Inline detector using reflective optics. Courtesy: ESRF, Instrument Support Group.

lens focal length. Objectives designed for infinity are usually not interchangeable with finite optical tube length and vice versa because the infinity lenses suffer from enhanced spherical aberration when used on a finite system. It is possible to use finite objectives with an infinity-corrected microscope but the magnification will be decreased and the spatial resolution degraded. Equally, the mixing of objectives and tube lenses from different manufacturers will result in a change of magnification and a loss of resolution.

10.5.1.1 Lens coupling: radiation damage  Recent research trends at synchrotrons show a move to use fast microtomography at increasingly higher X-ray energies, which poses a severe risk for the optical elements used in the microscope. At high energies, most of the X-ray photons will be transmitted by the converter screen and impinge on the first lens (see Fig. 10.2). Experiments at 65 keV, have resulted in severe radiation damage of a refractive objective already after 70 s. Therefore, a new design based on reflected light has been considered to avoid darkening of lenses by the X-rays (Ham et al., 2002). Two ways are possible.

The first uses a mirror between the scintillator and the microscope objective that reflects the visible light emitted by the luminescent screen at 90° to the X-ray beam and both prevents radiation damage of the optical components and facilitates shielding of the CCD camera. The mirror has to be placed sufficiently far away from the scintillator screen to avoid backscattered X-rays falling onto the screen. This setup requires a high mechanical precision of the mirror support that must be fixed at 45° to preserve a constant distance between the plane of the first lens and the plane object. An angular misalignment of the mirror will be a source of resolution loss as a result of a defocalization. The plane mirror is therefore only conceivable with low magnification.

The second solution to avoid radiation damage and consequent darkening of the optics is to use reflective optics (Di Michiel et al., 2005). Reflecting objectives for microscopes are generally simpler than refractive objectives since they consist of only two mirrors in the Schwarzschild configuration. The system is depicted in Fig. 10.6. It can be understood as an inverse Cassegrain system, such as a telescope. The system consists of a concave primary mirror and a con-
vex secondary mirror. Commercial reflecting objectives have a magnification not lower than $15 \times$. The present application requires a magnification of $5 \times$. Therefore, a specific development was made by Nachat$^{11}$ and the ESRF to produce a $5 \times$, NA=0.2 objective. The radius of the primary mirror is 135 mm, the radius of the secondary mirror is 50 mm and the two mirrors are separated by a distance of 85 mm. The long working distance of 90 mm has allowed a beam stopper to be fixed in front of the secondary mirror and thus protect the CCD and photo-eyepiece against radiation damage. In this design, the ESRF uses the infinity-corrected system. This solution has allowed to be maintained compatible with Olympus optics, and intermediate optical components can also be accommodated. No additional optics are needed to correct the image in the ‘infinity space’ between the objective and the tube lens. Advantages compared to refractive optics are: no scattering from the $45^\circ$ plane mirror in the folded detector system, and no damage to the optical components due to the protection by the beam stop. The disadvantages are the dimensions (96 mm diameter × 140 mm length), weight and cost.

10.5.2 Fibre-optical coupling
Fibre-optical coupling is largely preferred for demagnifying systems when a large FOV is needed. However, FO coupling has its own technical challenges. The tapered fibre optic must be attached to the CCD sensor with extreme precision. The optical coupling is usually a permanent bond with a thin layer of gels or epoxy optical cement that must withstand the heat cycle between room temperature and the CCD operation range of $-40^\circ$C to $-20^\circ$C. In addition to the hazardous operation of optical coupling, the general uniformity is degraded by spot blemishes (burned or broken fibres) (Ponchut, 2006) and line blemishes, called chickenwire (Gruner et al., 2002). Geometrical distortions owing to gaps and irregularities as an example of a lateral displacement of straight line for a multitaper CCD system (shear distortion at the taper’s edge) are an inherent disadvantage of a FO. The usable demagnification ratio is limited to less than 3 because the transmission efficiency decreases with increasing demagnification (see Fig. 10.7). Applications that require the largest FOV consist of a modular array of CCDs each with a tapered fibre optic. Figure 10.8 shows the CCD camera and its taper used on the medical beamline at the ESRF.

10.6 Readout based on CCD cameras
In the previous sections we have discussed the converter screens that convert X-ray photons to visible-light photons and the optical coupling that relays these optical photons to the photodetector. By far the most commonly used photodetector for tomography at synchrotron beamlines is the charge-coupled device (CCD) detector. In this section we will discuss some of the aspects of the CCD camera that are of importance to tomography experiments. Whereas spatial resolution and DQE are determined mainly by the converter screen and the optical

$^{11}$NACHET, 7 rue Ernst Chaput, 21059 DIJON, France; www.nachet.com.
coupling used, the frame rate and the dynamic range are mainly determined by the performance of the analogue-to-digital converter inside the CCD camera. Various sizes of CCD chips are commercially available, ranging from $1k\times1k$ to $4k\times4k$. At low frame rates, e.g., 1 frame per second (fps), these cameras allow for very large dynamic range images to be collected. High frame rates (25 fps) needed for fast and time-resolved tomography experiments are possible at the expense of dynamic range (10 to 12 bits). Figure 10.9 shows the present situation of scientific cameras used for tomography at synchrotrons.

10.6.1 Categories of CCD cameras
CCD cameras used for imaging experiments at synchrotrons can be roughly divided into three categories. The first category is the slow-scan camera, which approaches 16 true bit depth. This large dynamic range is achieved by slow and accurate analogue-to-digital converters in order to reduce readout noise combined with cooling of the CCD chip, which reduces the dark current (see below). Because of the relatively long readout times these systems are suitable for experiments with long exposure times (Wang et al., 2001). The second category are the fast 12-bit cameras. A large range of 12-bit CCD cameras with different pixel resolutions is commercially available, and they are used for many different applications (Jung et al., 2002). These cameras offer 30 to 60 fps and are used for fast tomography applications whenever speed is more important than dynamic range. The third category, which is the most popular for tomography applications at synchrotrons, is intermediate between fast and slow cameras and
Readout based on CCD cameras

Fig. 10.8. CCD camera with a taper coupling. Courtesy: ESRF, Instrument Support Group.

Fig. 10.9. CCD cameras used for tomography applications with synchrotron radiation.

represents a compromise between frame rate and dynamic range. Examples are the FReLoN2k (Bravin et al., 2003) and the pixel vision camera (Stampanoni et al., 2002), both based on the ATMEL TH7899M CCD chip. Another example is the Apogee system, see Tab. 10.1, that uses the Kodak KAF-1602 E chip (Ham et al., 2002, Beckmann, 2001).
10.6.2 Noise factors in CCD cameras

In general, one can categorize noise as either external or internal. External noise includes radiated electromagnetic interference and 50 or 60 Hz noise from the power supply. Internal noise, associated with the acquisition of an image, comprises readout noise, dark-current noise and photon shot noise. The readout noise depends on the readout speed (the faster the readout rate, the higher the noise) and the electronics controlling the CCD. This is why high-speed cameras have a reduced dynamic range. The readout electronics inside the FReLoN camera is specifically designed to reduce readout noise to a minimum, so as to allow for fast frame rates with high (or medium) dynamic range. CCD cameras are integrating detectors that also integrate the thermally generated noise, known as dark-current noise or dark noise. Inside the silicon of the CCD chip thermal energy will create electron–hole pairs. The electrons will be captured in the pixel wells and contribute to the signal. The dark current is highly temperature dependent, hence the CCDs are cooled to $-40^\circ \text{C}$ to $-20^\circ \text{C}$ to reduce the dark current, thus allowing for long exposure times. Cooling is achieved by thermoelectrical cooling (Peltier elements), forced air, water or liquid-nitrogen cooling. Since the accumulated dark noise increases linearly with time this factor is less important for fast tomography applications. The photon shot noise is a fundamental property of the quantum nature of light. According to statistical theory the number of visible photons collected by the CCD will exhibit a Poisson distribution. The charge induced by photon shot noise will also be Poisson distributed. The magnitude of the noise equals the square root of the number of corresponding signal photons.

10.6.3 Readout schemes of CCD cameras

The CCD detector allows for a certain flexibility for the readout of the pixels. For instance pixels can be binned together within rows or columns, increasing the frame rate at the cost of spatial resolution. Selected readout of a region of interest also increases the frame rate, but at the cost of image size (or FOV). Depending on the application one of the following general readout schemes is used: full-frame mode, frame-transfer mode, or pipeline mode.

The full-frame mode is the standard mode of operation of CCD detectors. In this mode the full chip is used to record the image. During readout the image is shifted down row-by-row to a horizontal register that is subsequently read out and digitized pixel-by-pixel through a single readout port. One can speed up the readout by using two or four readout ports (one at each corner). The speed-limiting step is, in general, the readout and digitization of the pixels, and not the shifting of the charge inside the pixel towards the readout node, which can be very fast. A disadvantage of the full-frame mode is the fact that during readout the CCD is still sensitive, while the image is shifted down, resulting in smearing of the image. Therefore, a shutter has to be placed in the X-ray path (X-ray shutter) or in the optical path, between the luminescent screen and the CCD, in order to stop image acquisition during readout.
An alternative mode of operation that circumvents this problem is the frame-transfer mode. In this mode, only half of the CCD chip is used to record the image, the other half of the chip is used as storage. While the image area is exposed to light, the storage area can be read. Once the reading is finished, the data in the image area is quickly shifted into the storage area for readout. Since this only involves shifting of charges and no digitization, this can be done very fast and no additional shutter is needed. In this mode, the CCD camera can run continuously (no dead time) with high frame rates. The disadvantage is that only half the CCD chip is used for recording data, thereby reducing the FOV.

Finally, there is the kinetic pipeline mode that is used for fast tomography. This can be considered an extreme case of the frame-transfer mode. This time, only a few lines are exposed to light, while the rest of the chip is masked. The exposed lines are shifted to the two nearest readout nodes and are read out in synchronization with the sample movement. In order to accelerate the image-acquisition frequency, the size of the exposed area should be as small as possible. The exposed area has to be equal to $2^n$ pixels, where $n$ is between 0 and 11 for a CCD with 2048 lines. A minimum dead time of 850 $\mu$s between the acquisitions of two lines can be achieved in this mode, thus allowing very fast tomography experiments.

10.7 Review of potential solutions for large field of view detector

The main part of this chapter has dealt with very high spatial resolution tomography with a small FOV. Another important field is tomography of larger objects, needing a detector with a much larger FOV and smaller spatial resolution. In this section, we will discuss two systems that show good potential to solve the needs for this particular field. The first system is a large flat panel based on amorphous silicon (a-Si) and developed for medical imaging. The second system is a flat panel based on complementary metal-oxide semiconductor (CMOS) technology.

10.7.1 a-Si-based flat panels

The medical imaging industry has recently introduced large fully digital flat-panel imagers that will gradually replace the traditional X-ray film cassettes in radiographic tables for general radiographic applications in vascular, thoracic and skeletal diagnostics. These flat-panel X-ray detectors are based on large-area solid-state integrated circuit technology in amorphous silicon. The integrated circuits consist of a photodiode array and an active TFT (thin-film transistor) matrix readout. A converter layer is deposited on top of the photodiode array. This layer can either be a direct converter such as amorphous Se or a scintillator converting X-rays to visible light such as cesium iodide (CsI). Trixell\textsuperscript{12} produces flat-panel imagers based on a-Si and using a CsI scintillator. A schematic layout of their system is given in Fig. 10.10. Their Pixium 4700 detector has an active

\textsuperscript{12}TRIXELL, 460 rue du Pommarin, 38430 Moirans, France; www.trixell.com.
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Fig. 10.10. Digital flat-panel detector for real-time X-ray imaging, TriXell. Courtesy of Trixell/Thales.

matrix of $30 \times 40$ cm, containing a total of $2480 \times 1910$ pixels each of $154 \, \mu m$. The data are digitized by 14-bit analogue-to-digital converters. At full resolution using all 5 megapixels a frame rate of 7.5 fps is obtained, 60 fps is possible with 4×2 binning. The CsI converter layer consists of many thin, rod-shaped, CsI crystal needles aligned parallel to one another, see Fig. 10.11(c). When an X-ray photon is absorbed by a CsI crystal the scintillator produces light that is reflected within the needle and transmitted out of one end of the needle without lateral diffusion. The photodiodes in the a-Si panel detect the scintillator light and convert it into an electrical signal. The primary benefit of CsI technology is the excellent DQE. The advent of these flat-panel digital imagers is revolutionary for medical imaging, but also offers opportunities for large FOV applications in the 30 to 100 keV energy range at synchrotron radiation sources. One disadvantage of the a-Si flat panels is the limitation in spatial resolution that amounts to $3.2 \, \text{LP/mm}$ for the pixium 4700 owing to the properties of the a-Si semiconductor material.

10.7.2 CMOS photodiode arrays

The second system we discuss here is produced by Hamamatsu Corporation. Their flat panels are based on the CMOS technology and also consist of a photodiode array but with smaller pixel size than the TFT arrays presented above. These imagers are very well suited for mammography and non-destructive analysis, but are certainly also interesting for many synchrotron radiation experiments (Yagi et al., 2004b). The C7930DP has $4416 \times 3520$ pixels at a $50 \, \mu m$ pitch ($220.8 \, \text{mm} \times 176 \, \text{mm}$) and can be run at frame rates of 1 fps (no binning) to 3.5 fps (2×2 binning), with 14-bit dynamic range (of the ADC). The spatial resolution obtained with this system is $10 \, \text{LP/mm}$. The C7930DP is designed with a scintillator as the converter layer. Two types of scintillating materials are available: CsI and Gadox ($\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$) which are deposited on aluminium, amorphous carbon and fibre-optic plates. As for the Trixel flat-panel imager the CSI scintillator is
Fig. 10.11. Difference between powder screen (a), transparent thin layer (b) and structured scintillator (c). In a powder screen, the light emitted leaves the screen by scattering in all directions. Scattering by neighbouring particles produces the familiar Lambertian angular distribution, a thicker screen involves a greater spreading. In structured scintillators, light emitted is propagated within the needle without significant spreading. In a transparent screen only the part of the light that is emitted at angles below the critical angle leaves the screen. The rest of the light is guided to the side of the screen by multiple reflection.

grown in parallel needles with a 6 to 7 \( \mu \)m diameter, resulting in the excellent spatial resolution, see Fig. 10.11(c).

10.8 Summary

In this chapter we have discussed the components of the detectors used for tomography at synchrotron beamlines. The main application is microtomography requiring detectors with high spatial resolution in the micrometre range. This is achieved by using very thin converter screens that are coupled via microscope optics to a CCD detector. We have shown the limitations of the various components of the system. The achieved spatial resolution of 0.5 \( \mu \)m is close to the theoretical limit of such a system. Therefore, the main gain that can still be achieved is in the quantum efficiency by using converter screens with higher X-ray stopping powers (higher-Z elements) and in the speed by using faster CCD detectors (or detection schemes). We have also presented the large field-of-view detectors developed by the medical imaging industry. These detectors will certainly be of interest for synchrotron experiments when the highest spatial resolution is not needed, but when speed and size are of importance.
10.9 References


and Methods in Physics Research A, 531, 75.


