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A single-photon CCD-based setup for in situ measurement of the X-ray spectrum of mammographic units

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ABSTRACT

A technique enabling in situ measurements of the spectrum of X-ray tubes employed in mammographic screenings is described. The technique involves the use of a commercially available CCD camera and a set of metal foils and is particularly useful to perform a fast evaluation of the spectral properties of a Mo anode mammographic system operating at standard flux levels. A description of the detector calibration procedure is first given, followed by a discussion of the study and choice of an appropriate set of X-ray attenuation foils. Finally, the use of the system for a spectroscopic characterization of a mammographic system is reported.

1. Introduction

X-ray mammography is currently the main established technique for the early detection of breast cancer. In the past few years, digital mammographic units have been spreading worldwide [1–3], due to their better performances that gives a significant dose reduction when compared to conventional screen-film devices [4]. However, for both the digital and the screen-film detection techniques, the energy spectrum of the X-ray tubes (including their filtering) is of a great concern in the performance evaluation of mammographic systems, demanding a careful and accurate quality controls in clinical environment [5,6]. Since the early 1990s several authors demonstrated the existence of an optimum spectral range, from 17 to 24 keV, for mammographic tests [7,8]. In fact, quasi-monochromatic X-rays give a better contrast in the case of breast lesions and enable the dose delivered to the patient to be reduced [9,10]. For these reasons, a lot of work has been done by several authors, and is currently going on, aiming to enable a fast investigation, possibly in situ, of the spectral properties of mammographic systems used in clinical diagnostics. In details, a relatively simple technique such as the attenuation analysis [11,12] has been initially employed in order to retrieve experimentally the spectrum of X-ray beams used for medical imaging purposes. After that, the use of an X-ray spectrometer based on Laue diffraction from a quartz crystal coupled to a CCD has been considered and successfully applied [13,14]. An alternative and more recent approach involves the use of solid-state detectors in the single-photon counting regime. As an example, Si photodiodes have been first employed to the purpose [15,16]. However, by far the most currently employed systems (in some cases commercially available) used for such a kind of measurements are based on cooled Ge [17–19] or, in the last years, on CZT or CdTe detectors [20–22]. It should be noted here that a lot of efforts are currently dedicated to increase the readout speed of solid-state detectors, in view of their extensive use for X-ray imaging [23–25] as well as for increasing their performances (in terms of statistics of the detected photons and number of pile-up events) for spectroscopic purposes [21,26].

As it is well known, due to their intrinsic linearity and to the large number of pixels, CCD detectors allow the spectrum of the impinging X-ray radiation to be retrieved without the use of external dispersing devices. This feature, originally exploited in astronomy [27], makes CCDs the most currently used X-ray detectors in different fields of fundamental and applied research [28], including, e.g., plasma physics [29–32], time-resolved X-ray diffraction [33], medical imaging [34] and nuclear physics [35,36].

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In contrast to the other detectors currently adopted for the characterization of X-ray tubes, the use of CCD detectors for single-photon spectroscopy relies upon the simultaneous detection of a large number of single-photon events (due to the large number of pixels) rather than to a high frequency sampling, thus relaxing the demand for a fast readout electronics. On the other hand, the single-photon condition, which is assured for CZT or CdTe detectors by pinhole collimators [21], is of course more difficult to be achieved in this case, due to the need of a suitable filtering of the CCD chip area, assuring a detection efficiency as uniform as possible over the whole spectrum of interest.

In this paper we report on the setup, characterization and use of an X-ray tube spectroscopic system based upon a CCD operating in the single-photon regime and a suitable set of X-ray filters allowing the system to be used directly after the X-ray window (i.e., in the space immediately before the breast support).

In the following section, the CCD detector used in our study will be described and its energy calibration by means of radioactive sources will be presented. The choice of a proper set of X-ray attenuators to make the system suitable for in situ characterization of mammographic tubes will be then discussed. Finally, in Section 3 a measurement on a tube operating at the S. Chiara Hospital in Pisa will be reported and discussed.

2. Description and characterization of the system

2.1. The CCD detector

The detector used in this work has been assembled by DTA Scientific Instruments (Pisa, Italy) using a KAF 0261E chip produced by the Eastman Kodak Company. This is a front-illuminated full-frame, buried channel CCD with 512×512 pixels with $20 \times 20\,\mu m^2$ size. The CCD is provided with a 16 bit ADC. In order to reduce the thermal noise, the chip, about 1 cm^2 of overall size, was kept, during the acquisitions considered in this work, at a temperature of $-5\,^{\circ}C$ by means of a Peltier device. An acquisition time of 5 s was used during the test of our system on a mammographic X-ray unit, which will be reported on later. This time was only dictated by the need of a manual triggering of the detector (due to the lack of an external output signal in the specific mammographic system employed) and was definitely greater than the X-ray pulse duration, as the single-photon condition was assured in our case by the use of a suitable attenuator set, which will be described later. In these experimental conditions the histogram of the background images had a FWHM of about $w_B \simeq 13$ ADU, resulting in a lower limit of about 150 eV to the final energy resolution. The sensitive area was shielded from low-energy and visible photons by a 10 μm thick Be window, whose transmissivity has been taken into account when processing the spectra. In order to correct for the dark charge pattern of the CCD, each image was processed by background subtraction using an unexposed image acquired under identical detector setting conditions.

The Quantum Efficiency (QE) of the CCD has been estimated using the data for the photon attenuation length $\lambda(E)$ in Si [37]. Indeed, assuming that all the photons absorbed in the Si substrate are detected, the efficiency can be written as $QE(E) = 1 - \exp(-t/\lambda(E))$ where t is the thickness of the active layer of the CCD. Although the parameter t was not known in our case, it can be easily found that for energies greater than about 7 keV and for typical values of t (5–10 μm [38]) one gets $t/\lambda(E) \ll 1$, so that the quantity $t/\lambda(E)$ is a good approximation, within about 10%, to the above formula. Thus, since the factor t is unimportant for our purposes (relative measurement), we used the function $1/\lambda(E)$ to take into account the energy dependence of the instrument efficiency.

As it is known, the pulse height distribution corresponding to a monochromatic X-ray signal acquired by a CCD detector is affected by a number of different "low energy side" components, due to the different possible physical processes occurring after the primary photon interaction (see, for instance, Ref. [39] and references therein). As an example, the final collected charge after the readout process can be considerably different even for events due to the same primary photon energy, as it is the result of the primary charge cloud drift, diffusion and recombination across the CCD [40,41]. Furthermore, the primary photon interaction in partially depleted pixel regions or in the channel stops can result in a reduction of the produced charge as well as in its splitting among neighbouring pixels [42,43]. Due to these reasons, dedicated algorithms are usually employed in order to retrieve the pulse height distribution and the spectrum of the impinging radiation. These algorithms usually start by distinguishing pixels with photon contribution from noise by means of a suitable rejection threshold, which was chosen as $3w_B$ in our case. In order to have the highest possible energy resolution, only single-pixel events (i.e., photon interactions whose resulting charge was collected in one pixel) are considered in the work described in this paper.

The detector was preliminarily calibrated in energy by using characteristic lines from ^{241}Am, ^{55}Fe and ^{133}Ba radioactive sources and the K_α line from a Mo anode X-ray tube. For each of the three radioactive sources 500 images were acquired, with an acquisition time of 30 s per image (only dictated by the activity of the radioactive sources used), keeping the source 5 cm away from the CCD chip. These parameters resulted in a total number of about 10^5 photons acquired for each source. As an example, Fig. 1 shows the pulse height distribution for the ^{241}Am source. The main lines coming from the ^{241}Am radioactive decay products are visible in the figure, up to an ADC level of about 2500, corresponding to a photon energy of about 25 keV (see below). As it is known, a strong decreasing of the detector efficiency occurs above this energy.

The response linearity of our CCD can be checked by looking at Fig. 2(a), showing the photon energy versus ADC level for the following lines: Np $L_{\alpha 1}$ ($E = 13.95$ keV) from the ^{241}Am source, Mn K_α ($E = 5.89$ keV) from the ^{55}Fe source, Cs $L_{\alpha 1}$ ($E = 4.29$ keV) and $L_{\beta 1}$ ($E = 4.62$ keV) from the ^{133}Ba source, Mo K_α ($E = 17.44$ keV) from the X-ray tube.

The relative energy resolution of the detector in our conditions as a function of the photon energy is shown in Fig. 2(b) (where δE is the FWHM of the observed peak). These values should be

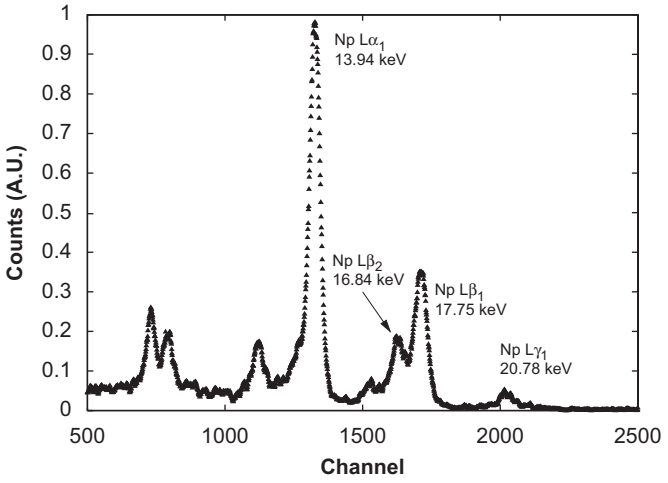

Fig. 1. Pulse height distribution for the ^{241}Am radioactive source.

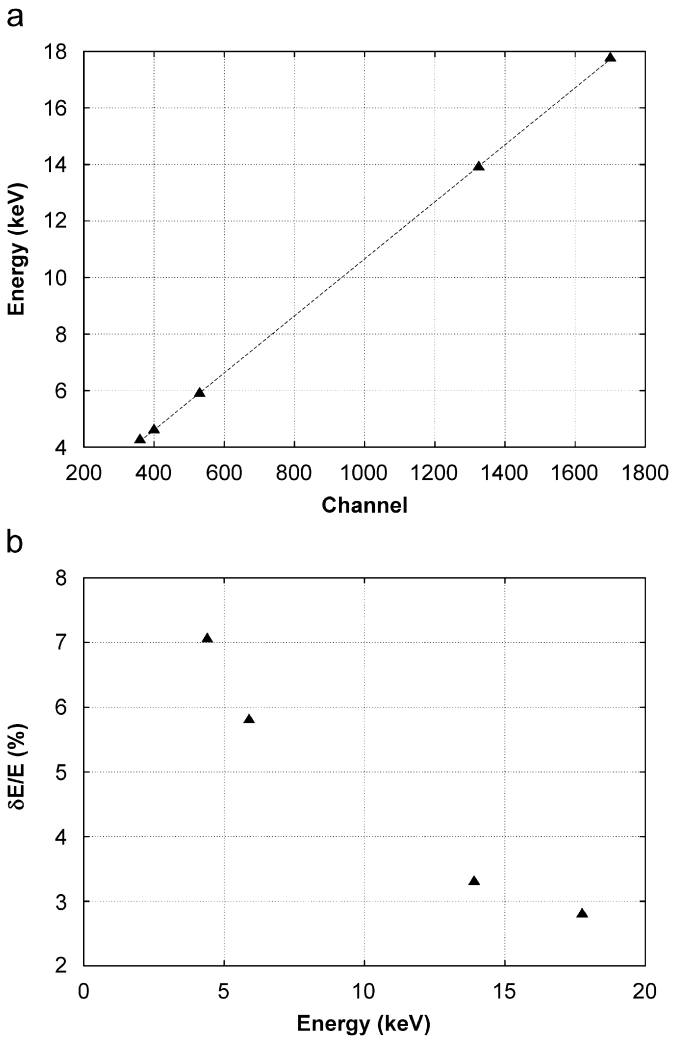

Fig. 2. (a) Photon energy versus ADC level for some emission lines in the range of interest for mammographic X-ray tubes. A fit with the function $E(\mathrm{keV}) = a + b \cdot \mathrm{ADU}$ is also shown. (b) Estimated energy resolution as a function of the photon energy.

compared with the expression $\delta E_{\mathrm{theor}} \simeq 2.355 w \sqrt{r^2 + FE/w}$, which can be found by theoretical considerations concerning the charge generation process in Si detectors (see, for example, Ref. [44]). In this formula $w \simeq 3.68$ and $F \simeq 0.15$ are the mean electron–hole pair creation energy and the Fano factor, respectively, [45], $r \simeq 14.59$ is the rms system noise in electrons (which includes readout and dark current noise) and E is the photon energy. A simple comparison shows that the energy resolution obtained with our detector is a factor 2 worse than the theoretical one provided by this formula. Similar (or slightly worse) values for the energy resolution with CdTe and CZT detectors have been recently reported in [21]. Better values have instead been obtained using crystal spectrometers in Laue configuration [14].

2.2. The attenuator set

As anticipated in the Introduction, CdTe or CZT detectors allow the in situ characterization of mammography X-ray tubes to be obtained provided that suitable pinhole collimators are used to reduce the direct X-ray flux (thus reducing the pile-up events to an acceptable level) and to avoid scattered radiation. In our case, due to the need of a wide-area attenuator, we considered the use of a suitable set of attenuating foils. The final goal was to achieve an attenuation as uniform as possible over the entire energy range of interest, namely from about 10 keV up to about 25 keV. This is a fundamental prerequisite, as in our case a strong variation of the attenuation versus photon energy (e.g., as occurring in the simple case of a single attenuating foil) would lead to an unacceptable statistical error, due to the single-photon counting, in some spectral ranges. At the same time, the presence of the absorption edges of the used materials had to be carefully taken into account, as it can lead, in general, to unacceptable abrupt variations in the transmission factor. This last issue can be addressed by a suitable choice of different materials with 'matched' edge energies, allowing both the edge effect to be reduced and an almost uniform transmission to be obtained, with no significant low energy cut-off in that specific energy range.

In detail, for a given set of N_{f} different foils (that is, materials), we looked for the best thicknesses $t_1, \ldots, t_{N_{\mathrm{f}}}$ by minimizing the quantity

$$\phi^2(t_1, \ldots, t_{N_{\mathrm{f}}}) = \sum_{j=j_{\min}}^{j_{\max}} [T(E_j; t_1, \ldots, t_{N_{\mathrm{f}}})\mathrm{QE}(E_j) - C]^2 \quad (1)$$

where the sum is performed over a discrete set of energy points E_j, with $E_{\min} = E_{j_{\min}}$, $E_{\max} = E_{j_{\max}}$. $T(E_j; t_1, \ldots, t_{N_{\mathrm{f}}})$ and $\mathrm{QE}(E_j)$ are the transmission factor for the given set of foils and the detector QE, respectively, evaluated at the energy E_j (the QE was estimated as reported above). C is the required, constant level of attenuation, which was estimated to be of the order of 10^{-4} for the conditions (distance, voltage and current) typical of the mammographic systems currently employed. The transmission factor T can be written as

$$T(E_j; t_1, \ldots, t_{N_{\mathrm{f}}}) = \exp\left(-\sum_{\alpha=1}^{N_{\mathrm{f}}} \frac{t_\alpha}{\lambda_\alpha(E_j)}\right)$$

where $\lambda_\alpha(E_j)$ is the attenuation length of the α-th material at the energy E_j.

The minimum of $\phi^2(t_1, \ldots, t_{N_{\mathrm{f}}})$ was searched, starting from different sets of foil numbers (N_{f}) and materials, by using a simplex minimization algorithm (see, for example, Ref. [46]). At the end, the final set of attenuators was chosen having in mind a compromise between the optimum condition, the commercially available materials and thicknesses and the requirement for a system as affordable as possible. The following six foils have been used: 12 μm thick Ti, 25 μm thick Y, 15 μm thick Mo, 5 μm thick Ta, 5 μm thick Pt, 10 μm thick Bi. Fig. 3 shows the transmission factor for this set of foils. The 'optimum' transmission factor is also shown, defined as the one corresponding to the optimum thicknesses as provided by minimizing formula (1). Also, Fig. 3 shows the product of our foil set transmission factor and the CCD QE. A convolution of the latter quantity with a Gaussian function accounting for the detector resolution, as evaluated in Section 2.1, is also shown. This curve was used for the final correction of the retrieved spectra.

3. In situ spectroscopy of a mammographic tube

The spectroscopic system described in the last section was first used to characterize an X-ray tube employed at the S. Chiara Hospital in Pisa (Italy) for mammographic screenings. The system (GIOTTO Hi-Tech produced by IMS) is equipped with a Mo anode, filtered with 0.76 mm thick Be and 0.03 mm Mo, and was operated in our case at a voltage $V = 22$ kV and with $I = 4$ mAs. Our CCD detector was placed on the standard breast support, that is at about 60 cm from the output window. The filter set was instead placed close to this window, that is as far as possible from the

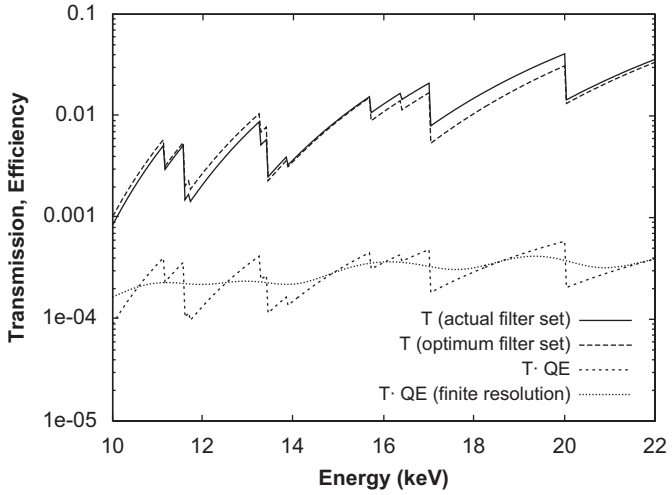

Fig. 3. Transmission factor T as a function of energy for the foil set used as attenuators. Both the curves corresponding to the actual used thicknesses and to an 'optimal' set of thicknesses (see text) are shown. The product $T \cdot \mathrm{QE}$ is also shown, together with its convolution with a Gaussian function accounting for the detector resolution as retrieved in Section 2.1.

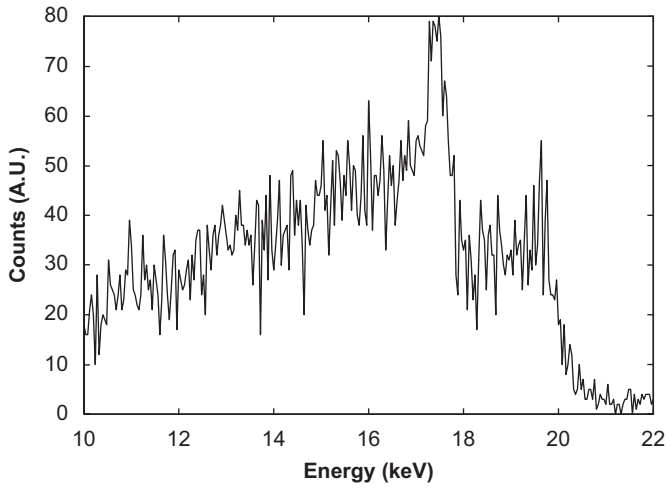

Fig. 4. Spectrum of the radiation emitted by the GIOTTO system, as obtained by our system with no corrections for the filter transmission and/or CCD efficiency.

detector, in order to minimize the fluorescence radiation on the detector. We stress here the fact that, unlike the systems based upon CdTe or CZT detectors, our system does not need any alignment procedure, thus relaxing the need for alignment systems and eliminating the possible effects on the retrieved spectrum due to the penumbra of the pinhole collimator [21].

Fig. 4 shows the raw spectrum obtained by summing up the contributions from 20 X-ray pulses from the tube with the above parameters. As it is well known, the typical X-ray pulse duration from standard tubes currently employed for medical diagnostics is of the order of a fraction of a second, depending on the current and on the manufacturing features. As already said in Section 2, for each X-ray pulse an acquisition was made with our detector with an acquisition time of 5 s, thus definitely greater than the pulse duration. The spectrum shown in Fig. 4 is the one as obtained from the raw counts with no correction for the filter transmission and/or the QE. The figure thus clearly shows the effectiveness of our filter set choice, as a nearly uniform number of photons (compatibly, of course, with the impinging X-ray radiation spectrum) was detected over the energy range of interest. Fig. 5 shows instead the final spectrum, as retrieved

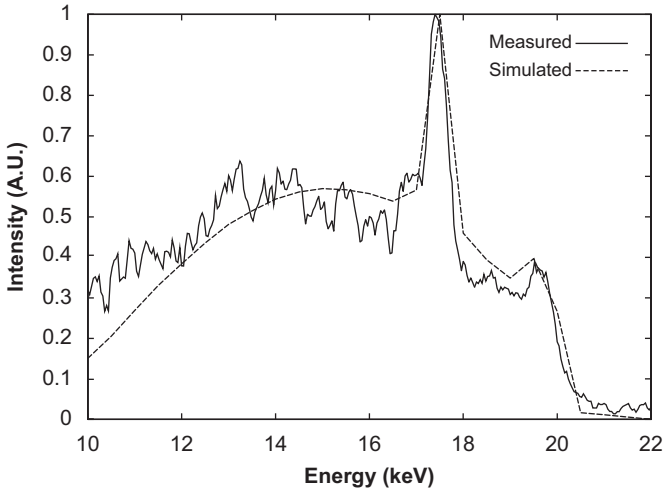

Fig. 5. Emission spectrum of a GIOTTO Hi-Tech mammographic system in the conditions given in the text, as obtained by our system. The corresponding spectrum calculated with the routine given in [47] is also shown.

from the one in Fig. 4 corrected for the filter transmission and the CCD QE as explained in Section 2.2. A 'theoretical' spectrum, calculated using the numerical routine given in Ref. [47] (which is based upon the work reported in Ref. [19]) is also plotted there, after correction for the Be and Mo filters in our mammographic system. The figure shows a very good agreement between the calculated and the measured spectrum. As a final comment on this measurement, we note here that it was performed using, as anticipated above, 20 X-ray tube pulses, for a total number of about 4×10^4 photons detected as single-pixel events. However, we point out that presently available large area, multi mega-pixel CCD detectors can significantly increase the number of simultaneous detected photons with respect to our case, hence enabling a single-shot measurement.

4. Summary and conclusions

We have reported on the setup and a first test of a spectroscopic system for the in situ measurement of the spectral properties of mammographic X-ray units. The system is based upon a CCD detector operating in the single-photon regime and a set of ad hoc X-ray attenuating foils, allowing the system to be used just above the breast support with no structural manipulation of the mammographic unit. A calibration of the detector has been preliminarily performed by using radioactive sources and an X-ray tube. The attainable energy resolution is similar to the one recently obtained with CdTe or CZT detectors used for the same purpose. A set of six foils of different materials was studied by using a dedicated algorithm, in order to get a detected number of photons as uniform as possible over the energy range of interest, in the operational conditions usually encountered for mammographic examinations. The system can thus be used for in situ characterization of mammographic systems in clinical environments, taking advantage of its feature of requiring no particular alignment procedures. A first measurement has been reported here, carried out at the S. Chiara Hospital in Pisa (Italy), assessing its validity as a fast, easy-to-use system for quality controls.

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