

(19) World Intellectual Property Organization
International Bureau



(43) International Publication Date
19 April 2001 (19.04.2001)

PCT

(10) International Publication Number
WO 01/27656 A1

(51) International Patent Classification⁷: G01T 1/24, 1/29

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(21) International Application Number: PCT/SE00/01954

(22) International Filing Date: 9 October 2000 (09.10.2000)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
9903655-0 8 October 1999 (08.10.1999) SE
60/158,465 8 October 1999 (08.10.1999) US

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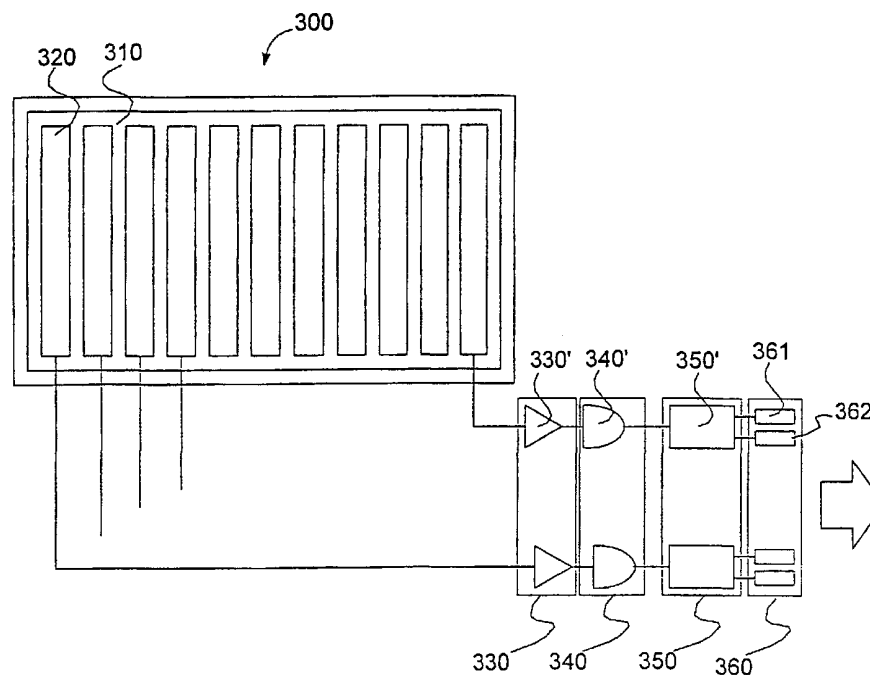
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(81) Designated States (national): AE, AG, AL, AM, AT, AT (utility model), AU, AZ, BA, BB, BG, BR, BY, BZ, CA, CH, CN, CR, CU, CZ, CZ (utility model), DE, DE (utility model), DK, DK (utility model), DM, DZ, EE, EE (utility model), ES, FI, FI (utility model), GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KP, KR, KR (utility model), KZ, LC, LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW, MX, MZ, NO, NZ, PL, PT, RO, RU, SD, SE, SG, SI, SK, SK (utility model), SL, TJ, TM, TR, TT, TZ, UA, UG, US, UZ, VN, YU, ZA, ZW.

(84) Designated States (regional): ARIPO patent (GH, GM, KE, LS, MW, MZ, SD, SL, SZ, TZ, UG, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, CY, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, GW, ML, MR, NE, SN, TD, TG).

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(54) Title: METHOD AND ARRANGEMENT RELATING TO X-RAY IMAGING



(57) Abstract: The present invention relates to a method of detecting a number of photons in an x-ray detecting arrangement comprising a number of spaced apart sensors, said detected photons indirectly creating an amount of free charges proportional to the photon energy. The method comprises weighting said photons by means of signals with respect to possible photon charge-share between at least two adjacent sensors.



WO 01/27656 A1

**Published:**

- *With international search report.*
- *Before the expiration of the time limit for amending the claims and to be republished in the event of receipt of amendments.*

For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.

Title

METHOD AND ARRANGEMENT RELATING TO X-RAY IMAGING

5 TECHNICAL FIELD OF THE INVENTION

The present invention relates to a method and arrangement of detecting a number of photons in an x-ray detecting arrangement comprising a number of spaced apart sensors, said detected photons indirectly creating an amount of free charges proportional to the photon energy.

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BACKGROUND OF THE INVENTION

The possibility of an x-ray imaging was discovered a long time ago, but imaging systems of today still use the following trivial principle. Photons coming from a spectrum of energies are
15 passed through an object and then detected. If a massive body, like a tumour, is included in this object, a shadow (fewer photons) will fall on the detector and an image is created. Of course, there are stochastic fluctuations in the number of photons that pass the tissues. A sufficient amount of radiation is needed to make these fluctuations small relative to the difference in expected number photons passing the different tissues. It is desirable to make the most efficient use of the detected
20 photons in order to minimize the patient dose. Better detectors have been applied for this purpose, and less radiation containing information about the object escapes undetected. However, very little attentions have been drawn to the possibilities to increase the efficiency using information about the energy spectrum after the object. Digital systems open up new possibilities concerning this idea.

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US 5,665,969 describes an x-ray detector, designed to operate as an imaging spectrometer for imaging of a subject. The x-ray detector measures energy of individual x-ray photons in each of a plurality of pixels in the x-ray detector. The pixels of the x-ray detector are readout at a rate such that the likelihood of arrive of more than one x-ray photon in each pixel during a readout period is
30 negligible. Because x-ray photons with different energy levels will create different magnitude responses in the x-ray detector, the measurements made by the x-ray detector can be weighted according to the energy level of the detected x-ray photons. Thus, responses due to noise or x-ray photons which contribute little or no x-ray attenuation information can be discarded or weighted

to eliminate or reduce their effect on any resulting image. Conversely, measurements due to x-ray photons which provide significant attenuation information can be weighted significantly.

According to this patent, the optimal energy weighting one should use is the theoretical optimal one, that is approximately proportional to the negative third power of energy. There is nothing mentioned about charge shares and the optimal weight curve which can be used in reality. To be able to weight the photons with respect to the information content in a realistic way for semiconductor detectors, the signal sharing between the detector pixels must be considered. This has not been obvious until now. US '969 has not considered nor thought of this possibility. If charge sharing is used in this case for energy weighting, the resultant image will be deteriorated, specially in case of mammography. The suggested method works for detectors having large pixels, in which charge sharing can be neglected and hence this is not a problem. In case of small detectors with spacial resolution the charge sharing will affect the output signal.

In traditional detectors, the signal is usually integrated for each pixel and each individual photon is not considered. In these types of detectors, normally used in hospitals and x-ray examination of material, the charge sharing between the pixels is positive, which increases the signal quality. It is also inherited in the detector structure which is not considered in signal processing. However, for photon-counting detectors, the signal sharing causes problems.

Examples of detectors, on which the present invention can be applied to are disclosed in US 4,937,453 and the pending Swedish Patent Application No. 9900856-7. US 4,937,453 discloses a method and apparatus for detecting x-ray radiation in a radiographic imaging context using so-called "edge-on" detectors. It is particularly useful in conjunction with slit and slot scan radiography. In accordance with this invention, detectors are constructed and arranged such that substantially all of the energy from an X-ray to be detected is discharged in the detector. In this way a detector is provided which provides a direct electronic read out, high X-ray stopping power and high spatial resolution while obtaining good signal collection efficiency without the use of excessively high voltage levels. In the preferred embodiment, solid state x-ray detectors are constructed such that the thickness of the detector along the direction of incident x-rays is long enough that substantially all of the x-ray energy is discharged in the detector.

The application No. 9900856-7 refers to a method of obtaining improved radiographic images

consisting of orienting a semiconductor radiation detector. The orienting step comprises a selection of an acute angel between the direction of incident radiation and a side of the detector such that the incident radiation mainly hits the side.

- 5 Fig. 1 is a schematic illustration of a detector 100 comprising a semiconducting substrate 110 and spatially arranged sensor or electrode strips 120.

Common for these detectors is that stripes of sensors are arranged spaced from each other on a silicon substrate and the x-rays incident onto both the sensors and the space between them.

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SUMMARY OF THE INVENTION

It is one objective of the present invention to enhance the prior art methods by means of a simple but yet efficient arrangement.

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Another object of the present invention is to provide a novel method and arrangement for detecting and analysing x-rays in an efficient and accurate manner.

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Yet another object of the present invention is to reduce the effects of charge sharing and trapping in a photon charge detector arrangement.

For these reasons the method of the invention comprises weighting said photons by means of signals with respect to possible photon charge-share between at least two adjacent sensors.

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The method comprises the further step of disregarding the smaller signal or adding together two signals when signals from the sensors appear within a small time window on at least two neighbouring strips.

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The method comprises the further step of providing an optimal weighting curve by calculating distributions of charge sharing for each energy bin in a photon spectrum that enters the detector and using an optimal theoretical weight curve for said spectrum. Moreover, for each energy bin of the incoming spectrum, it is calculated how a large fraction of photons that will be recorded in preferably all different bins in a recorded spectrum and how many photons that are not counted

and the distribution of signals in the neighbouring sensor. The method comprises the further step of calculating the distribution of real photon energies that belong to a recorded energy bin, wherein said weight belongs to a certain bin being the convolution of said distribution and theoretical weight factors.

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In one embodiment a weight curve for trapped photon charges is computed, which further comprising a distribution of signals, e.g. through simulation, from each energy bin in the photon spectrum, which also includes trapping, and through backwards calculation, calculating for each bin in a detected pulse amplitude spectrum how the photons which contributed to the bin are distributed in its energy spectrum.

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In a preferred embodiment the method further comprises the steps of irradiating an object through an x-ray source, detecting the beam having a spatial object information and a spectrum filtered by the object by means of said sensors, wherein the signal from the sensors is a signal deriving from a detected photon, signal from each x-ray detector for each sensor being readout and an amplitude of the signal being compared to a threshold level, generating an output for signals above or below said threshold value as an output for an amount of time, which relates to the time the signal is above or below the threshold levels, and if the signal from an x-ray photon is shared between two sensors and triggers an adjacent comparator, the two comparators outputs generating a simultaneous signal. The method comprises the further step of detecting said simultaneous signal and initiating a charge sharing, indicating if the amplitude of a signal is high or low, for indicating if the x-ray photon is high energy or low energy, and counting the number of photons through said indication representing spectrums of signal amplitudes for each image pixel.

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A preferred arrangement for detecting and counting photons in an x-ray beam includes a detector arrangement comprising sensors, and at least to each sensor coupled amplification unit, comparator means, logic unit and counter. Preferably, the counter comprises a first and a second counter for each sensor and the first and second counters correspond to high and low energy photons, respectively. Moreover, the comparator is connected to a threshold value and if the signal exceeds a predetermined threshold level the output of a comparator is a logical signal as output for an amount of time, which relates to the time the signal exceeds the threshold levels. If the signal from an x-ray photon is shared between two sensors and two adjacent comparators, the signal generated is same at simultaneously. Additionally, simultaneous signals are detected by the logic

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units and if at least two adjacent comparators have same signals charge sharing is indicated. Moreover, it includes interface means for connection to further processing means.

BRIEF DESCRIPTION OF THE DRAWINGS

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The invention will in the following be further described in a non-limiting way with reference to the accompanying drawings in which:

Fig. 1 schematically illustrates a detector arrangement for detecting x-rays,

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Fig. 2 is an example of a weight curve,

Fig. 3 is a schematic block diagram of a detector arrangement embodiment, according to the present invention, and

Fig. 4 is a block diagram illustrating the steps of a method according to the present invention.

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DETAILED DESCRIPTION OF EMBODIMENTS

The difference in absorption probability between the normal tissue and the tumour becomes larger with decreasing photon energy. Therefore, a low energy photon carries more contrast information and should be given a higher statistical weight after detection. Mathematically, it can be shown
20 that the photons should be given statistical weights that are approximately proportional to the negative third power of their energy.

Beside the examples of detectors disclosed above, also gaseous detector, such as a Parallel Plate Chamber where the gas volume is oriented edge-on to the incident X-rays can be employed.

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Imaging systems are normally integrating, that is, the signals created from individual photons in the detector are added together. Since the signals are proportional to the photon energy, the statistical weights differ four powers of the energy from the optimal weighting. In digital photon counting systems, all photons are given unity statistical weights, which is therefore a more
30 efficient way of obtaining information describing the object. The optimal way would be to use a detector with high energy resolution, and to register the detected spectrum. The energy bins in this spectrum can then be added together in the optimal weighting manner. A simplifying compromise to this method is to record the spectrum into only a few bins. An extreme case of this is to use

only two energy bins, each one with an associated weight factor (dual weighting).

In reality, the spectrum that would be registered is not the correct one that is entering the detector. The reason is the following. The photons that are detected indirectly create an amount of free charges that is proportional to the photon energy. These charges are collected on electrodes, and the created signal is proportional to the number of charges collected. This is what is used in an energy sensitive detector system where the signals are sorted into a spectrum according to their sizes. When a photon is transformed into free charges, they flow towards electrodes. Ideally, all free charges of the type that are collected (electrons or holes) are collected at the nearest electrode (sensor) strip, so it contributes in the right image pixel. In general however, an amount of charges will be shared with the strip that is the second nearest to the interaction (charge sharing). Yet, another amount of free charge will become trapped inside the detector volume, and will not reach the electrode in time to contribute to the signal (trapping). The trapping effects are significant between the electrodes (same as charge sharing), which depends on the flow of free charges in an area between the electrodes and immediately under the surface of the electrodes. In the mentioned area the electrical potential drop is affected by the charges in the oxide layer between the electrodes. Consequently, the electrical drop is low and the free charges are slowed down and trapped. However, outside the electrodes the trapping effect is insignificant. Theoretically, it is possible to disregard the trapping, but trapping depends on the electrode geometry, specially the distance between the electrodes exposed to the oxide layer, bias voltage on the detector and manufacturing.

The charge sharing occurs in both silicon and gas microstrip electrodes (and others).

The effects of these two mechanisms are that the photons in general will be registered having less energy than they really had, and subsequently, they will be given to large statistical weights. The readout electronic has a threshold to discriminate real signals from noise (mainly appearing in the pre-amplifier). This threshold level is well below the least energetic photons in the spectrum entering the detector, but, because of the charge losses, some photons will give rise to signals below this threshold and will not be registered at all. In addition, charges that are shared will be interpreted as a low energy photon. Moreover, the trapped charges will partly be scattered in time and partly be all too few to be able to pass the threshold. This is of course a drawback, since it is a false photon that will acquire a large weight.

One possible way to reduce these problems is to use anti coincidence when reading out the detector signal, i.e. when signals appear within a small time window on two neighbouring strips, the smaller signal is disregarded. Another way would be to add the two signals together, and thereby reconstruct the initial photon energy. It is not possible to fully reconstruct the right spectrum in this way. For example, the charge lost by trapping is not considered. The trapping depends on many factors, such as detector material, material purity, the sensor or electrode width, the space between the sensors etc.

This distortion of the spectrum suggests that the theoretical optimal weighting calculated before is not optimal in reality. One significant difference is that the lowest end of the spectrum should have low statistical weight instead of highest, since a large fraction derives from false photons (if anti-coincidence is not used).

The optimal weighting curve can be found after calculating the distributions of charge sharing for each energy bin in the photon spectrum that enters the detector. It is desirable to use the optimal theoretical weight curve for this spectrum, although this is not the spectrum that is recorded. For each energy bin of the incoming spectrum, it is calculated how a large fraction of photons that will be recorded in all the different bins in the recorded spectrum, how many photons that were never counted at all (because their signals were reduced below the electronic threshold level), and the distribution of signals in the neighbouring strip. Then it is possible to calculate the distribution of real photon energies that belongs to a recorded energy bin. The weight belonging to a certain bin is then the convolution of this distribution and the old theoretical weight factors. The fraction of false photons is given zero weight. The fractions are normalized, so that the sum of fractions in the real spectrum plus the fraction of false photons are considered equal. The fractions of undetected photons with energy corresponding to the considered bin are then added to the convolution after being multiplied with the corresponding weight factor.

Fig.2 is an example of an optimal weight curve for a 5-mm tumour when anti-coincidence has not been used. To be compared with E^{-3} according to prior art.

In a preferred embodiment shown in fig. 3, an arrangement according to the invention comprises a set of detector arrays 300, comprising sensors 310, which are connected to an amplification block 330, a comparator block 340, a logic block 350 and a counter block 360.

Referring to the block diagram of fig. 4, the operative steps according to the present invention in conjunction with fig. 3 are as follow: an x-ray source irradiates 410 an object. The beam from the x-ray source has a spectrum characteristic of the source. The beam is collimated 420 or refracted onto an object to be examined. The beam filtered by the object 430 obtains a spatial object
5 information and a spectrum, and incident 440 onto the detectors. The signal from the detectors is a signal deriving from a detected photon. The signal shape depends on the photon energy and conversion position in the detector. One photon can produce signals in more than one channel due to the charge sharing.

10 The signal from each x-ray detector 300 for each channel, i.e. each sensor 310, is readout 450 by first being amplified by an amplifier 330', after the amplifier the amplitude of the signal is compared to threshold levels in a comparator 340'. If the signal exceeds a predetermined threshold level, the output of comparators is sat, e.g. by means of a logic one, as an output for an amount of time, which related to the time the signal exceeds the threshold levels. If the signal from an x-ray
15 photon is shared between two channels and triggers an adjacent comparator, these two comparators will signal ("1") at the same time. To determine a low or high energy photon, there are preferably two threshold values, a first low and a second high value. A low energy photon is detected if the signal exceeds the low threshold but not the high threshold. A high energy photon threshold is detected if both low and high thresholds are exceeded.

20 The simultaneous signals will be detected by the logic units 350' following the comparators, and if an adjacent comparator is "1" a flag will be set for charge sharing. This can be achieved by standard logic such an AND-gate for signals from adjacent comparators. There can be another flag set indicating if the amplitude is high or low, i.e. if the x-ray photon is high energy or low
25 energy.

After the logic block 350, there are the counters 360; in a simple embodiment there is one counter 361 for high energy photons and one counter 362 for low energy photons. Alternatively, it is possible to use several counters if more accurate measurement of the energy is wanted. The
30 normal case is that there is not a charge-sharing phenomenon and this flag is not set and the counter corresponding to the energy of the photon is incremented. If the charge sharing flag is set only the counter for one channel is updated, which could be any of the two channels. However, it is important that two channels are not incremented at the same time. In this case only the counter

corresponding to the highest photon energy is updated, because of charge sharing the energy is not known and the weighting a high energy photon as a low energy photon is avoided since this severely degrades the DQE. Weighting a few low energy photons as high energy photons is not so severe.

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The content of the counters, which are spectrums of signal amplitudes for each image pixel, can then be readout 460, e.g. through a shift register or similar, by a computer for storing data, making energy weighting with optimal weight-function, image processing and presentation. Compensation for charge losses and charge sharing in the detector is included in the optimal weight function.

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Obviously, above described arrangement is given as an example and other arrangements, e.g. including A/D-converters, microprocessors etc., can occur.

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With respect to the trapping, the trapped charges will be handled as the "false" electrons in the charge sharing case, i.e. the share of the trapped charges is counted and this share is weighted zero. However, it is not possible to discriminate the trapped charges using anti-coincidence. As a result of the trapping the signal from an electrode will decrease irrespective of the fact that it is effected by the charge sharing or not. The weight curve for trapping is computed in same way as above. Firstly, a distribution of signals, e.g. through simulation, from each energy bin in the photon spectrum is computed, which also includes trapping. Then through backwards calculation it is calculated for each bin in the detected pulse amplitude spectrum how the photons which contributed to the bin are distributed in its energy spectrum.

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New optimal weighting curves have been calculated taken into account charge sharing, and achieved efficiency for different weighting methods have been determined based on simulations. In the following the results of simulations of 100 and 50 micron pitch detectors are disclosed. Charge sharing is included, but not trapping. The efficiencies are expressed as DQE (Detective Quantum Efficiency). The reference is the signal to noise ratio (SNR) squared in an ideal detector without charge losses and using the theoretical optimal weighting curve. The DQE depends on the object that is imaged, and in the simulations, a 40 mm thick breast with a 5 mm tumour and a 250 micron calcification were modelled. In reality, of course, it is not possible to choose the weighting curve that is optimal for the tissue in a particular pixel. Optimal weight curves are, however, not

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that different. Preferably, it is best to choose the curve that corresponds to the most difficult tissue (and significant for diagnosing purpose) to detect in an image.

In addition, the DQE depends on the electronic threshold level. The results disclosed below are those for the threshold level that gives the highest DQE, that is, optimal threshold levels have also been determined in this study. A 30 keV tungsten spectrum was used in the simulations. In all weighting methods disclosed in tables 1-3 below, except for the "optimal" case as disclosed in the tables, a signal corresponding to less than 14 keV was weighed as a 14 keV photon. This improves the efficiencies since it is known that no photon below 14 keV passes unabsorbed through the object. The optimal weighting, however, assigns the best individual weights also for these lowest energy bins. The uncertainties in the DQEs stated below are ~1% and the threshold levels are within 200-300 electrons.

Table 1 discloses DQEs and optimal threshold levels for different weighting methods in the case where no anti-coincidence is used. The results for a detector with 100 micron pitch and for 50 micron pitch, respectively, are shown in respective column. The threshold levels should be multiplied with 3.6/1000 to convert from electrons to keV.

Table 1

Weighting method	DQE for tumour		DQE for calcification		Optimal threshold level for tumour (electrons)		Optimal threshold level for calcification (electrons)	
micron pitch	100	50	100	50	100	50	100	50
Old theoretical	0.76	0.64	0.80	0.67	4000	4000	4000	4000
Counting	0.77	0.72	0.80	0.75	2800	2800	2800	2800
Integrating	0.70	0.69	0.72	0.72	2400	2000	2400	2000
Dual weighting	0.82	0.75	0.85	0.78	3200	3200	3200	3200
Optimal	0.86	0.79	0.87	0.81	2400	2400	2400	2400

The DQEs dropped from 1 to 0.76 and 0.80 for tumour and calcification respectively when the old theoretical weighting curve was used and charge sharing was introduced. The efficiency becomes worse than for a photon counting system. The two weighting bins have been optimized in the case of dual weighting. If the signal corresponds to an energy higher than 22-23 keV, then it should be given a weight that is 0.61 or 0.68 for 100 and 50 micron pitch, respectively, in the case of tumour

and 0.64 and 0.71 in the case of calcification. Otherwise, the weight is unity.

Table 2 shows DQEs and optimal threshold levels for different weighting methods in the case when anti-coincidence is used. The results for a detector with 100 micron pitch and for 50 micron pitch, respectively, are shown in separate columns. The threshold levels should be multiplied with 3.6/1000 to convert from electrons to keV.

Table 2

Weighting method	DQE for tumour		DQE for calcification		Optimal threshold level for tumour (electrons)		Optimal threshold level for calcification (electrons)	
micron pitch	100	50	100	50	100	50	100	50
Old theoretical	0.79	0.8	0.84	0.82	2000	2000	2000	2000
Counting	0.81	0.8	0.84	0.85	2000	2000	2000	2000
Integrating	0.7	0.7	0.73	0.73	2000	2000	2000	2000
Dual weighting	0.88	0.9	0.9	0.88	2000	2000	2000	2000
Optimal	0.9	0.9	0.92	0.9	2000	2000	2000	2000

The cut in the spectrum should now be made at 22-23 keV in case of dual weighting and 100 micron pitch. The higher energy weight factors are 0.58 and 0.62. In case of 50 micron pitch, the cut should be at 21-22 keV and the weight factors 0.62 and 0.66. The threshold levels should know be set as low as possible without allowing false counts due to electronic noise. 2000 electrons were chosen, which allows all photons to be detected, even if the charge sharing is 50% (which should be considered correct when trapping is not considered.)

Table 3 includes DQEs and optimal threshold levels for different weighting methods in the case when coincident signals on two neighbouring strips are added. The results for a detector with 100 and 50 micron pitches, respectively, are in separate columns. The threshold levels should be multiplied with 3.6/1000 to convert from electrons to keV.

Table 3

Weighting method	DQE for tumour	DQE for calcification	Optimal threshold level for tumour (electrons)	Optimal threshold level for calcification (electrons)

micron pitch	100	50	100	50	100	50	100	50
Old theoretical	0.92	<i>0.91</i>	0.94	<i>0.9</i>	2000	<i>2000</i>	2000	<i>2000</i>
Counting	0.81	<i>0.82</i>	0.85	<i>0.8</i>	2000	<i>2000</i>	2000	<i>2000</i>
Integrating	0.69	<i>0.7</i>	0.74	<i>0.7</i>	2000	<i>2000</i>	2000	<i>2000</i>
Dual weighting	0.9	<i>0.89</i>	0.93	<i>0.9</i>	2000	<i>2000</i>	2000	<i>2000</i>
Optimal	0.94	<i>0.94</i>	0.95	<i>0.9</i>	2000	<i>2000</i>	2000	<i>2000</i>

The cut in the spectrum should now be made at 21-22 keV (100 micron pitch) and 18-19 keV and 19- 20 keV (50 micron, tumour and micro-calcification respectively) in case of dual weighting.

The higher energy weight factors are 0.57 and 0.61 (100 micron pitch) and 0.60 and 0.65 (50 micron pitch). Again, the threshold levels should know be set as low as possible. Accidental coincidence of independent photons has not been considered.

The reason why the optimal weighting is still a few percent better than the old theoretical, and the optimal method compensates for the fact that photons only detected on one strip in general creates lower signal than corresponds to its energy.

The invention is not limited the shown embodiments but can be varied in a number of ways without departing from the scope of the appended claims and the arrangement and the method can be implemented in various ways depending on application, functional units, needs and requirements etc.

CLAIMS

1. A method of detecting a number of photons in an x-ray detecting arrangement comprising a number of spaced apart sensors, said detected photons indirectly creating an amount of free
5 charges proportional to the photon energy,
characterised by
weighting said photons by means of signals with respect to possible photon charge-share between at least two adjacent sensors.
- 10 2. The method according to claim 1,
characterised by
disregarding a smaller signal, when signals from the sensors appear within a small time window on at least two neighbouring strips.
- 15 3. The method according to claim 1,
characterised by
adding two signals together when signals from the sensors appear within a small time window on at least two neighbouring strips.
- 20 4. The method according to claim 1,
characterised by
providing a weighting curve by calculating distributions of charge shares for each energy bin in a photon spectrum that enters the detector.
- 25 5. The method according to claim 4,
characterised by
using a theoretical weight curve for said spectrum.
- 30 6. The method according to claim 5,
characterised by
for each energy bin of the incoming spectrum, calculating how a large fraction of photons that will be recorded in preferably all different bins in a recorded spectrum and how many photons that are not counted and the distribution of signals in the neighbouring sensor.

7. The method according to claim 6,
characterised by

the further step of calculating the distribution of real photon energies that belong to a recorded energy bin.

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8. The method according to claim 7,
characterised by

said weight belonging to a certain bin being the convolution of said distribution and theoretical weight factors.

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9. The method according to claim 1,
characterised by

said a weight curve for trapped photon charges is computed.

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10. The method according to claim 9,
characterised by

- computing a distribution of signals, e.g., through simulation, from each energy bin in the photon spectrum, which also includes trapping, and
- through backwards calculation, calculating for each bin in a detected pulse amplitude spectrum how the photons, which contributed to the bin are distributed in its energy spectrum.

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11. The method according to any of previous claims,
characterised by

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- irradiating an object through an x-ray source,
- detecting the beam having a spatial object information and a spectrum filtered by the object by means of said sensors, wherein the signal from the sensors is a signal deriving from a detected photon,
- reading out a signal from each x-ray detector (300) for each sensor (310) and comparing

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- an amplitude of the signal to a threshold level,
- generating an output for signals above or below said threshold value as an output for an amount of time, which relates to the time the signal is above or below the threshold levels, and

- if the signal from an x-ray photon is shared between two sensors and triggers an adjacent comparator, the two comparators outputs generating a simultaneous signal.

12. The method according to claim 11,

5 *characterised by*

detecting said simultaneous signal and initiating a charge sharing.

13. The method according to claim 11,

characterised by

10 indicating if the amplitude of a signal is high or low, for indicating if the x-ray photon is high energy or low energy.

14. The method according to claim 12,

characterised by

15 counting the number of photons through said indication representing spectrums of signal amplitudes for each image pixel.

15. The method according to claim 12,

characterised by

20 if a charge sharing flag is set only a counter for one channel is updated, which could be any of the two channels, said counter corresponding to a highest photon energy being updated, so that weighting a high energy photon as a low energy photon is avoided.

16. An arrangement for detecting and counting photons in an x-ray beam according to any of

25 methods of preceding claims,

characterised in

that it includes a detector arrangement (300) comprising sensors (320), and at least to each sensor coupled amplification unit (330), comparator means (340), logic unit (350) and counter (360).

30 17. An arrangement according to claim 16,

characterised in

that said counter comprises a first and a second counter (361, 362) for each sensor.

18. An arrangement according to claim 16,
characterised in
that said first and second counters correspond to high and low energy photons, respectively.

5 19. An arrangement according to any of claims 16-18,
characterised in
that said comparator is connected to a threshold value.

10 20. An arrangement according to claim 19,
characterised in
that if the signal exceeds a predetermined threshold level the output of a comparator is a logical signal as output for an amount of time, which relates to the time the signal exceeds the threshold levels.

15 21. An arrangement according to claim 20,
characterised in
that if the signal from an x-ray photon is shared between two sensors and two adjacent comparators, the generated signal is same simultaneously.

20 22. An arrangement according to claim 21,
characterised in
that simultaneous signals are detected by the logic units and if at least two adjacent comparators have same signals charge sharing is indicated.

25 23. An arrangement according to any of claims 16-22
characterised in
that it includes interface means for connection to further processing means.

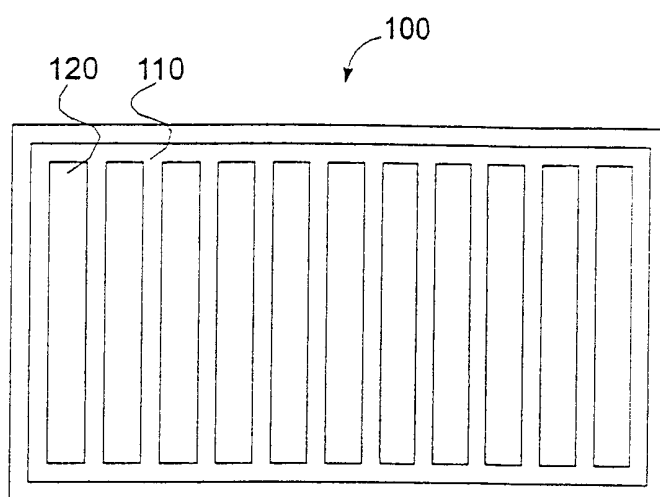


Fig. 1

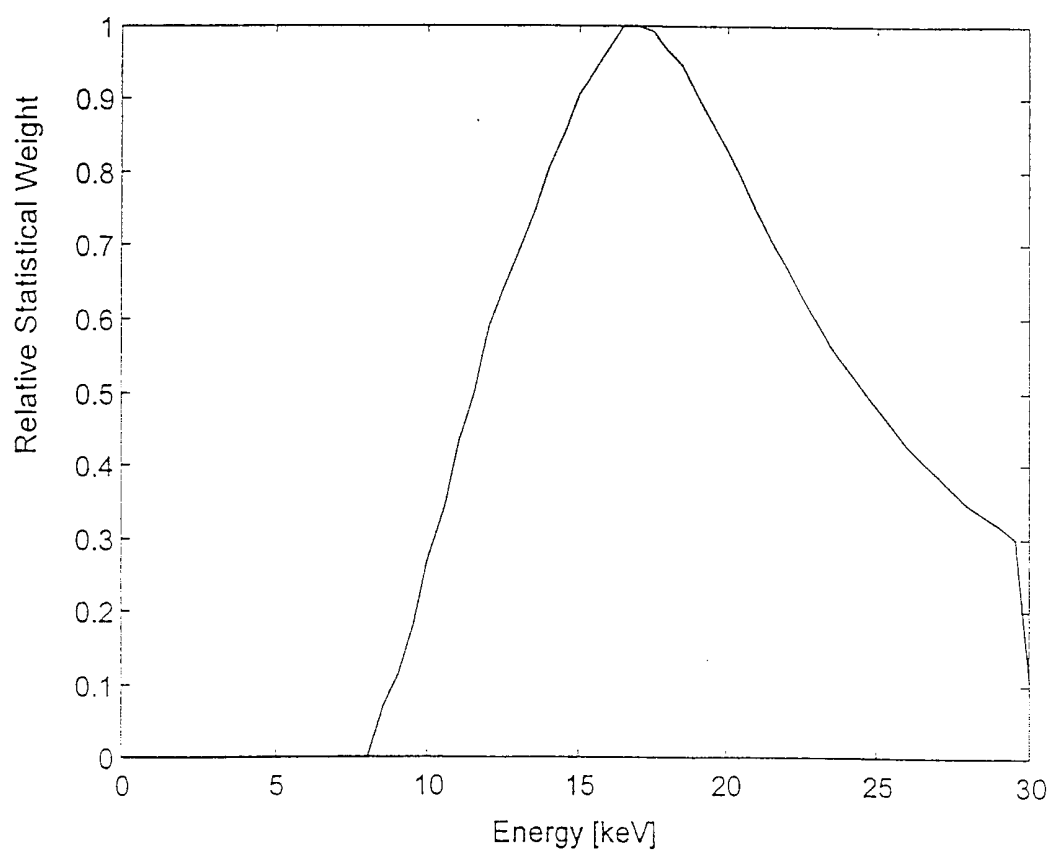


Fig. 2

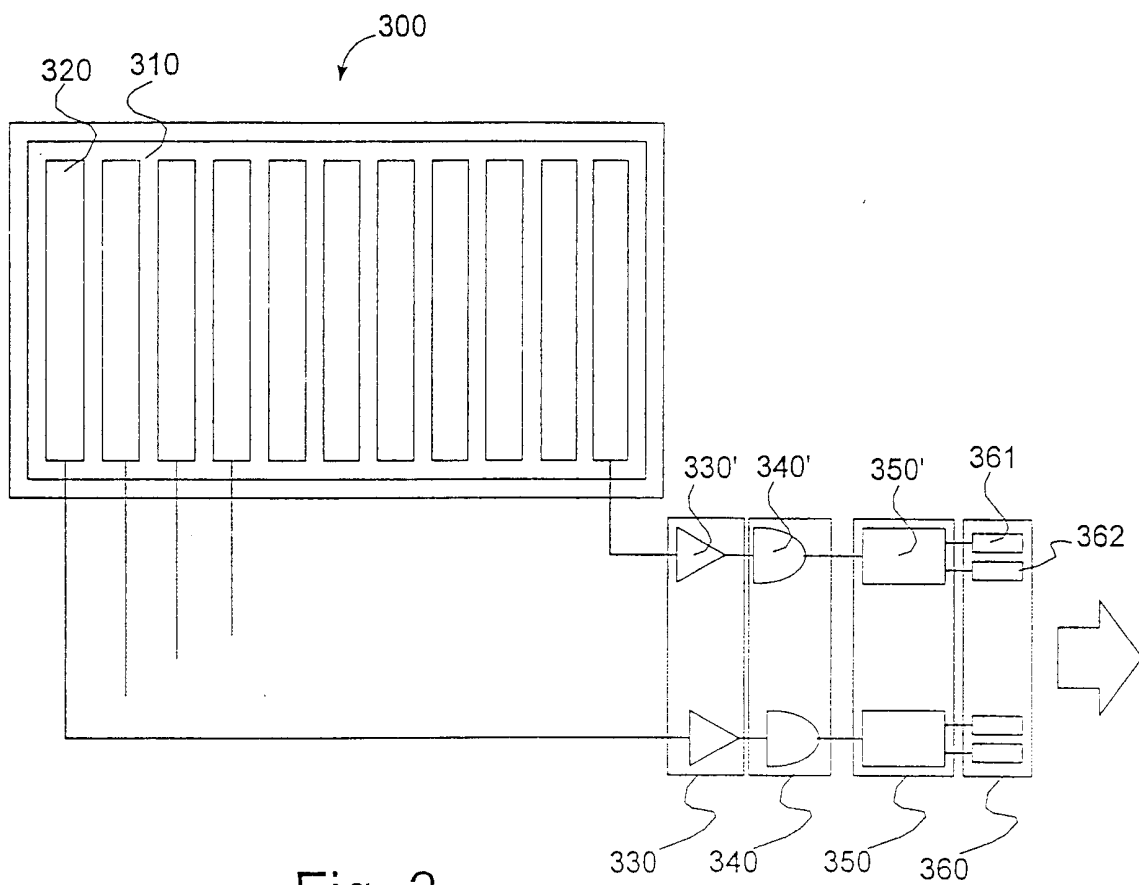


Fig. 3

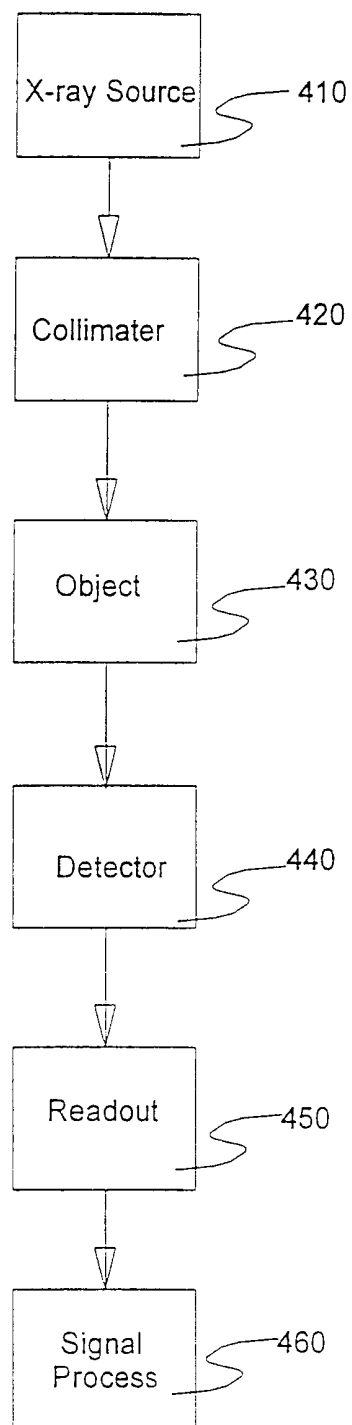


Fig. 4

INTERNATIONAL SEARCH REPORT

International application No.
PCT/SE 00/01954

A. CLASSIFICATION OF SUBJECT MATTER

IPC7: G01T 1/24, G01T 1/29

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC7: G01T, H01L

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

SE,DK,FI,NO classes as above

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

IEL, INSPEC

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	YIN, S. ET AL. "A LOW-DOSE HIGH CONTRAST DIGITAL MAMMOGRAPHY SYSTEM (DIGIMAM). Nunuclear Science Symposium, 1998. Conference Record. 1998 IEEE 8-14 NOV. 1998. pp. 1656-1659 vol. 3 --	1-23
X	US 5665969 A (J. U. BEUSCH), 9 Sept 1997 (09.09.97), column 2, line 12 - line 61; column 6, line 6 - line 13; column 8, line 52 - line 63, col. 9 line 40-49, col. 10 li. 55-65 col. 11 li. 6-50. fig. 4, 5 --	1-23
A	US 5481115 A (YU-LUNG HSIEH ET AL), 2 January 1996 (02.01.96), column 5, line 26 - line 59 --	1-23

☒ Further documents are listed in the continuation of Box C.☒ See patent family annex.

* Special categories of cited documents:

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"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"I" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance: the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance: the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search

Date of mailing of the international search report

6 February 2001

13 -02- 2001

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INTERNATIONAL SEARCH REPORT

International application No.

PCT/SE 00/01954

C (Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 4864140 A (W.L.ROGERS ET AL), 5 Sept 1989 (05.09.89), column 1, line 34 - line 47; column 5, line 58 - line 63 --	1-23
A	US 4937453 A (R.S. NELSON), 26 June 1990 (26.06.90) --	1-23
A	EP 0776126 A1 (PLANMED OY), 18 November 1996 (18.11.96) -- -----	1-23

INTERNATIONAL SEARCH REPORT

Information on patent family members

27/12/00

International application No.

PCT/SE 00/01954

Patent document cited in search report			Publication date	Patent family member(s)		Publication date
US	5665969	A	09/09/97	WO	9641213 A	19/12/96
US	5481115	A	02/01/96	DE	4406996 A	15/09/94
				US	5338936 A	16/08/94
				US	5479021 A	26/12/95
				EP	0526970 A	10/02/93
				JP	5302979 A	16/11/93
				US	5210421 A	11/05/93
US	4864140	A	05/09/89	NONE		
US	4937453	A	26/06/90	US	5017782 A	21/05/91
EP	0776126	A1	18/11/96	FI	99074 B,C	13/06/97
				FI	955599 D	00/00/00
				US	5872364 A	16/02/99