(54) Title: SYSTEM AND METHOD FOR PHOTON DETECTION

(57) Abstract: The present invention provides a detection system for measuring photon fluxes within a large dynamic intensity range. The detection system comprises at least one photo detector element (3) arranged to detect incident radiation and comprising at least one avalanche photo diode [APD]. The detection system comprises measuring means (5, 7, 9, 11, 15) for measuring the rate and magnitude of the discrete electric pulses, and the mean current, generated by the detector element (3) as a result of the incident radiation. The detection system is characterized in that it comprises a bias regulator (13) comprising means for altering a bias voltage (V_B) applied to the detector element (3) between at least a first and a second voltage level, said first voltage level being below the breakdown voltage of the detector element (3), and said second voltage level being above the breakdown voltage of the detector element (3).
with information concerning authorization of rectification
of an obvious mistake under Rule 91.1

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Title: Method and device for photon detection

Technical field

The present invention relates to a detection system for photon sensing and for measuring photon fluxes according to the preamble of claim 1, a scanner for computed tomography (CT), positron emission tomography (PET), single photon emission computed tomography (SPECT), or portal imaging applications, or any combination of these, according to the preamble of claim 20, use of a photon detection system in PET, SPECT, CT, or radiotherapy portal imaging applications, or any combination of these, and a method for photon sensing and for measuring photon fluxes according to the preamble of claim 22.

Background of the invention

Photon sensing is an important component in many medical and technical applications. Besides optical imaging, there are several types of applications concerned with detecting ionizing radiation where the radiation is converted into optical photons in scintillators and a large dynamic intensity range is needed, such as industrial radiography, nuclear safeguards, environmental radiation monitoring and airport security. Other important ionizing radiation detection applications are diagnostic and therapeutic medical imaging applications.

Molecular imaging is a rapidly growing modality in medical diagnostics and of particular interest is the recent development and rapid growth of combined computed tomography (CT) and positron emission tomography (PET) systems. The photomultiplier tubes (PMTs) used together with scintillator crystals to detect X-rays and gamma rays in current state-of-the-art PET scanners and single photon emission computed tomography (SPECT) scanners are about to be replaced with photo detectors based on semiconductor technology, such as avalanche photo diodes (APDs). A major rationale behind this ongoing transition to silicon-based devices is that they can operate in a strong magnetic field and, thus, provide the technology for combined CT/SPECT/PET systems and magnetic resonance imaging (MRI) systems.

In applications with low to moderate radiation fluxes, such as PET/SPECT applications, the radiation detectors are usually operated in a “photon counting mode” (pulse mode) which
means that each signal corresponding to a detected primary photon with the desired properties will generate a pulse that can be used by a data acquisition system, additionally after being processed by fast photon counting electronics, such as preamplifiers, discriminator circuits, etc. If the radiation detectors consist of APDs coupled to scintillator crystals, the APDs may be operated in either proportional mode or "Geiger mode". When operated in proportional mode, each APD generates pulses whose magnitudes are proportional to the number of detected photons generated in the scintillation process and hence proportional to the energy deposited by the primary radiation. Under such conditions the APDs are usually run at high gain (typically up to ~ 1000) in order to optimize the signal to noise ratio and the APD signals are passed on to preamplifier and discriminator circuits designed to select events depending on energy and timing. Geiger mode operation is achieved by operating the APDs above their breakdown voltage. When operated in Geiger mode, the internal gain is extremely high (approximately \(10^6\) - \(10^7\)) and the breakdown current caused by a diode avalanche has to be limited to a fixed value by a current limiter ("quenching resistor"). Thus, an APD operated in Geiger mode is advantageous when very high internal gain is needed but has the drawback of converting any input signal to a pulse that is independent of the signal amplitude above a certain threshold. Hence each channel of such a system can be regarded as a digital “yes or no” sensor. To overcome the above mentioned drawback of Geiger mode APDs, arrays of such detectors, sometimes called silicon photomultipliers (SiPMs) or multi-pixel photon counters (MPPCs), have been constructed. These arrays can emulate the linear response of normal photomultiplier tubes and APDs operated in proportional mode by adding the outputs from a large number of individual Geiger mode operated detector pixels.

In applications with high radiation fluxes, such as high-speed X-ray CT imaging or portal imaging systems, utilization of APDs operated in single-photon counting mode is not possible or inefficient due to saturation effects due to dead time and pulse pile-up. The maximum count rate per detector element in state-of-the-art single-photon counting systems is well below the mean count rate per unit area required in standard CT imaging. In photon counting mode, such as for PET, APDs can practically be used for rates up to about 10 MHz, mainly limited by the decay time of the scintillator. Therefore, common photodiodes that photovoltaically generate a current that is substantially proportional to the energy flux of the measured radiation are normally used for such high-dose rate applications.
Consequently, for applications requiring a wide dynamic range of photon fluxes, such as the aforementioned combined CT/PET system, two separate detector systems has previously been required; one detector system comprising, e.g., common photodiodes for enabling high-rate current readout needed for fast CT scanning, and one high gain detector system consisting of, for example, APDs for enabling high-resolution single-photon readout needed for PET (and SPECT) scanning. The requirement of two separate detector systems implies more complex, expensive and cumbersome radiation detection devices.

WO 2006/034585 A1 presents a method and system for acquiring CT images using a photon-counting detector, such as an avalanche photodiode (APD) or an array of APDs (e.g., a SiPM).

However, as aforementioned, CT scans often require a high radiation dose and detectors operating in photon-counting mode are unable to measure high photon fluxes due to saturation effects.

For example, the scientific article entitled “Architecture of a Dual-Modality, High-Resolution, Fully Digital Positron Emission Tomography/Computed Tomography (PET/CT) Scanner for Small Animal Imaging” by Fontaine, R. et al. published in IEEE Transactions on Nuclear Science, 52:691, 2005, discloses a detector consisting of a scintillator crystal coupled to an APD functioning only in the normal single-photon counting mode. Since the pulse counting approach is technically challenging at very high rates, the authors suggest limiting the X-ray flux such that it can be processed by the current state-of-the-art electronics. However, limits on the X-ray flux could compromise the CT image quality. Another drawback of the pulse-counting approach was described by the authors: “As the X-ray energy is one order of magnitude lower than the PET 511 keV radiation, the front-end analog electronics must be designed with a wide dynamic range and a short pulse decay time to accommodate the large amplitude range and high rate of detector signals.”

Recent developments have, however, shown that it may be possible to utilize semiconductor photodetectors, such as APDs, also for quantifying high radiation fluxes. For example, the scientific article entitled “The Road to the Common PET/CT Detector” by A. Nassalski et al., found on the Internet (http://www.ipj.gov.pl/anassalski/PLIKI/AN%20TNS%202006.pdf), 2007-02-16, disclosed detector consisting of a scintillator crystal coupled to an APD
functioning not only in the normal single-photon counting mode but also in a current mode. At low radiation intensities the photon counting mode was used and standard pulse signals were picked up from the APD. At high incident photon flux on the other hand, when operation in photon counting mode was compromised due to dead time and pile-up, a signal corresponding to the mean current of the APD, which is substantially proportional to the incident photon flux at the scintillator crystal, was picked up and suggested to be used to compensate for the non-linear count rate response at high photon rates. By using such a dual mode APD, the dynamic range of the detector was shown to be considerably widened (about a factor 20, limited by the strength of the radioactive source).

When operated in pulse mode and used together with a scintillator in medical X-ray imaging, such a dual mode APD can be used to detect single primary photons up to fluxes of millions of photons per mm$^2$ per second. When operated in current sensing mode the dynamic range can be extended by many orders of magnitude, in principle far beyond what is practically useful for most applications.

However, there are several problems associated with such an APD-based dual modality sensor system. First, the attainable energy resolution of the scintillator readout by the APD is limited by Groom’s “theorem”, as described in the scientific article "Silicon photodiode detection of bismuth germinate scintillation light" by Donald E. Groom in “Nuclear Instruments and Methods in Physics Research”, Volume 219, Issue 1, 1 January 1984, Pages 141-148.

Secondly, if the APD is operated in proportional mode (e.g. for PET), the internal gain of the detector is not sufficient and additional amplification is needed before the signals can be handled by the pulse processing electronics. If the APD on the other hand would be operated in Geiger mode, the output signal amplitude carries no information of the primary photon energy incident on the scintillator crystal. Such problems could be remedied by using a SiPM device comprising a plurality of APDs connected in parallel. However, if a SiPM is used and the number of secondary photons incident on the photo detector per unit area of the device, within the single-pixel recovery time (which for state-of-the-art devices lies in the range ~100ns – 1 μs), becomes comparable to the pixel density (currently limited to around 1000 – 2000 pixels per mm$^2$, depending on the type and manufacturer), significant nonlinearities in the sensor response will occur. In addition, the finite single-pixel recovery time of Geiger mode APDs makes them count rate limited and introduces non-linearities in the current response at high photon fluxes. Not only does this preclude quantification of high radiation
fluxes in single-photon counting mode due to saturation, the nonlinearities in the sensor response also severely compromise the readout of the continuous mean current in current mode operation.

Summary of the invention

It is an object of the present invention to provide a detection system and a method for measuring photon fluxes within a large dynamic intensity range.

The object is achieved by a detection system for photon sensing and for measuring photon fluxes comprising at least one photo detector element arranged to detect incident radiation and comprising at least one avalanche photo diode (APD). The detection system comprises measuring means for measuring the rate and magnitude of the discrete electric pulses, and the mean current, generated by the detector element as a result of the incident radiation. The detection system is characterized in that it comprises a bias regulator comprising means for altering a bias voltage applied to the detector element between at least a first and a second voltage level, said first voltage level being below the breakdown voltage of the detector element, and said second voltage level being above the breakdown voltage of the detector element.

By arranging a bias regulator to control the bias voltage of the detector element in the above described manner, the detector element can be operated in Geiger mode (bias voltage above the break-down voltage) at low to moderate photon fluxes requiring a high gain for optimizing the readout in pulse mode, and in proportional mode (bias voltage below the breakdown voltage) at high photon fluxes for optimizing the readout in current mode, thus extending the dynamic intensity range of the detector element.

The detector element preferably comprises a plurality of APDs, constituting what is sometimes called a silicon photomultiplier (SiPM) or a multi-pixel photon counter (MPPC). The large gain of the SiPM when operated in Geiger mode and the small pixel size yields a strong reduction of capacitive noise and a high immunity against noise pickup and avoids the need for expensive high-gain amplifiers. Furthermore, by adding the output from a plurality of APDs operated in Geiger mode the SiPM can emulate the linear response of normal
photomultiplier tubes and APDs operated in proportional mode. In addition, the limitation implied by Groom’s “theorem” is no longer valid.

Preferably, the bias regulator comprises an operator input device, allowing an operator of the detection system to switch between the two different bias voltage levels. Thereby, an operator can chose to optimize the readout in pulse mode for low to moderate rate applications, such as PET and SPECT, and to optimize the readout in current mode for high rate applications, such as standard CT.

The bias regulator may also be arranged to automatically switch between the two different bias voltage levels applied to the detector element in dependence of a parameter value indicative of the radiation intensity incident on the detector element. The bias regulator is then arranged to apply a bias voltage above the breakdown voltage of the detector element if the parameter value is indicative of a low incident radiation intensity, and a bias voltage below the breakdown voltage of the detector element if the parameter value is indicative of a high incident radiation intensity.

The parameter value received by the bias regulator and used in the decision of whether to apply a bias voltage above or below the breakdown voltage of the detector element may be a parameter value relating to the rate and magnitude of the discrete pulses and/or the mean current generated by the detector element as a result of the incident radiation. It may also be a parameter value relating to the radiation intensity $I_R$ emitted from a radiation source.

By providing a detector system being able to measure photon fluxes within a large dynamic radiation intensity range, the present invention eliminates the need for two separate detection systems in applications requiring such a large radiation intensity range.

State-of-the-art technology, e.g. for dual-modality CT/PET scanners, is based on the use of separate detection systems combined into one imaging system. The present invention improves both the cost efficiency and the performance of such a scanner by providing the means for a single detection system being able to operate in single photon counting Geiger mode for PET, SPECT or low-rate CT applications and in current mode for high-rate CT applications. The combined CT/PET/SPECT scanner according to the present invention can
provide greatly enhanced medical diagnostics such as in oncology as well as cardiology and neurology.

Furthermore, since the detection system of the present invention utilizes photo detectors based on semiconductor technology, the detection system can be integrated in MRI systems, thus providing multiple-modality CT/PET/MRI or CT/PET/SPECT/MRI systems.

One particularly interesting field of application of the present invention is for combined diagnostic and therapeutic imaging. This concerns imaging in connection with radiation-based cancer therapy where it is important to couple as closely as possible the diagnostic imaging of the patient with the portal imaging that is performed online to verify the dose delivery. The invention will enable the design of a single detector system that can perform both diagnostic CT/PET/SPECT as well as the portal imaging tasks during a cancer therapy session. This will greatly increase the accuracy, quality and efficiency of the dose delivery and lead to higher protection of healthy tissues by reducing uncertainties in patient positioning and by providing the possibility of online corrections to the therapeutic beam.

The object is also achieved by a method for photon sensing and for measuring photon fluxes within a large dynamic radiation range. The method preferably comprises the use of a detection system as disclosed above.

**Brief description of the figures**

Fig. 1 shows an embodiment of the detector system according to the present invention, realized in a system intended for combined PET/CT or PET/SPECT/CT scanning.

Fig. 2 shows the photo detector element used in the detector system shown in Fig. 1

Fig. 3 shows a flowchart illustrating an example of a method for measuring photon fluxes within a large dynamic range according to the present invention.

**Detailed description**
As described above, there are several types of applications concerned with measuring photon fluxes within a large dynamic range. Industrial radiography, nuclear safeguards, airport security, optical imaging and diagnostic and therapeutic medical imaging are just some examples of such applications. Although applicable to all of these exemplary fields of application, the present invention will henceforth be described in the context of a system for combined computed tomography (CT), positron emission tomography (PET), and single photon emission computed tomography (SPECT). CT, PET and SPECT are well-known medical structural and functional imaging techniques for obtaining three-dimensional images of the internals of an object and the fundamental principles of these techniques need not further be disclosed herein.

Fig. 1 shows part of a detection system for a combined PET/SPECT/CT scanner according to one embodiment of the present invention.

During operation in CT mode a radiation source 4 emits ionizing radiation, illustrated as arrows in the figure, towards a target region 8 with an intensity regulated by a radiation control unit (not shown) contained within or connected to the radiation source 4. The radiation passing through the target region 8, or secondary radiation arising from positron annihilation or directly from radioactive decays within the 8 (CT and PET/SPECT, respectively), is absorbed by scintillators 1 in the detector system.

Scintillators are well known in the art and serve to fluoresce photons at characteristic wavelengths in response to absorbed incident ionizing radiation. The number of generated fluorescence photons is substantially proportional to the energy of the incident primary X-ray or gamma photon. The scintillators 1 may be any known type of scintillators, such as organic crystal scintillators, organic liquid scintillators, organic plastic scintillators or inorganic crystal scintillators.

The scintillators 1 are optically coupled to a plurality of semiconductor photo detector elements 3. Optionally, the scintillators 1 may be coupled to the photo detector elements 3 via light guides (not shown). In Fig. 1, there is a one-to-one correspondence between detector elements 3 and scintillators 1, i.e. each photo detector element 3 is coupled to one scintillator. However, each photo detector element 3 may be coupled to more than one scintillator 1, or more than one photo detector element 3 may be coupled to each scintillator 1.
The photo detector elements 3 may comprise a single avalanche photo diode (APD) 2 or, preferably, a plurality (an array) of APDs 2 constituting what is known as a silicon photo multiplier (SiPM) device, multi-pixel photon counter (MPPC), or multi-cell avalanche photo diode (MAPD). This is illustrated schematically in Fig. 2. For the sake of simplicity an array of APDs will henceforth be referred to as a SiPM 10.

The photo detector elements 3 are in turn connected to measuring means for measuring the rate and magnitude of the discrete electric pulses, and the mean current, generated by the at least one APD 2 as a result of the incident radiation. Consequently, each photo detector element 3 is arranged to operate both in single photon counting mode (pulse mode) and current mode. The means for measuring the rate and magnitude of the electric pulses and the means for measuring the mean current may comprise any components and circuitries known in the art and need not to be described in detail. Preferably, the detector elements 3 are arranged to operate in pulse mode and current mode simultaneously, thus allowing simultaneous measurements of both the rate and magnitude of the discrete electric pulses, and the mean current, generated by the detector elements 3.

In this exemplary embodiment of the invention, each photo detector element 3 is coupled to a current processing circuitry 9 via a current-sensitive amplifier 5 for current-mode operation, and to a pulse processing circuitry 11 via an optional amplifier 7 for pulse-mode operation. It is also possible to add the output from several photo detector elements 3 and then pass on the summed signal to the amplifiers 5, 7 and the processing circuitries 9, 11, or to add the signals produced after amplification before passing them to the processing circuitries 9, 11. Appropriate current and pulse processing circuitries 9, 11 for CT, PET and SPECT applications are well known in the art and need not further be disclosed herein.

The above described components constitute a detector module 6 and a plurality of such detector modules 6 constitutes the detector system in the combined CT/PET/SPECT scanner. The total number of detector modules 6 used in the combined CT/PET/SPECT scanner may, of course, vary with the size of the area that needs to be covered by detectors and the position resolution requirements for the particular application.
In this embodiment, the current processing circuitry 9 and the pulse processing circuitry 11 of each detector module 6 are coupled to a data processing unit 15 via two separate channels CH 1, CH 2. In another embodiment of the invention, the output from the current processing circuitry 9 and/or the output from the pulse processing circuitry 11 of several detector modules 6 may be added before being sent to the data processing unit 15. That is, the data processing unit 15 may have a varying number of data channels. Many channels yield a good image resolution but increase the manufacturing cost and the size of the detection system due to the need for a larger amount of electronic components. The data processing unit 15 registers data corresponding to the energy and timing of the received signals and the mean currents induced in the photo detectors and uses this information in a known manner to create diagnostic images of the target region 8. In the field of medical imaging, the volume “seen” by the detector system in CT/PET/SPECT applications is often referred to as “the field of view”. The field of view does not necessarily coincide with the irradiated target region in CT applications and images are, of course, only possible to obtain for the field of view. The term “target region” used in this description should be interpreted as the field of view in any of the above mentioned medical imaging applications.

The photo detector elements 3 are also electrically connected to means 13 for providing a controlled bias voltage $V_B$ thereto. The means 13 for providing a controlled bias voltage $V_B$ to the detector assemblies 3 will henceforth be referred to as the bias regulator 13. The bias regulator 13 can in its simplest form be a voltage supply arranged to provide two different voltage levels to the photo detector elements 3: one level above the break-down voltage of the photodetectors 3, and one level below the break-down voltage of the photodetectors 3. In this case, the bias regulator 13 preferably comprises an operator input device for allowing an operator of the equipment to switch between the two levels. For example, an operator is then able to choose a voltage level above the breakdown voltage of the photo detector elements 3 for PET/SPECT or low to moderate rate CT applications, and a voltage level below the breakdown voltage of the photo detector elements 3 for high-rate CT applications.

In another embodiment of the invention, the bias regulator 13 comprises a voltage supply and a control unit for regulating the voltage $V_B$ supplied to the photo detector elements 3 in dependence of certain variables relating to the photon flux intensity $I_\lambda$ incident on the photo detector elements 3. In the illustrated embodiment, the bias regulator 13 is arranged to receive such radiation related information from the data processing unit 15, via a communication...
channel 17, and/or from the radiation source 4, via another communication channel 19. These communication channels 17, 19 are illustrated as dotted lines in Fig. 1. The radiation source 4 or its control unit may be arranged to provide the bias regulator 13 with information related to the radiation emitted from the radiation source 4, e.g. the emitted radiation intensity $I_R$, and the data processing unit 15 may be arranged to provide the bias regulator 13 with information related to the pulses or the mean current outputted by the detector modules 6. Of course, the radiation source 4 may be connected to the data processing unit 15, in which case information related to the radiation source 4 or its control unit can be provided to the bias regulator 13 via communication channel 17.

The detector system comprising a bias voltage $V_B$ regulatory system according to the present invention thus allows the photo detector elements 3 to be operated in Geiger mode by applying a bias voltage $V_B$ above the breakdown voltage and in normal mode by lowering the bias voltage $V_B$ below the breakdown voltage. Thereby, the detector system according to the invention can be used in applications requiring a large dynamic radiation range and be optimized for the radiation intensity currently used. By operating the detector elements 3 in Geiger mode, the internal gain of the detector elements 3 is sufficient to detect single photons, thus optimizing the detector system for low to moderate radiation rate applications, such as PET, and by operating the detector elements 3 in proportional mode, the mean current outputted from the detector elements 3 will stay proportional to the incident photon flux even at high fluxes, thus optimizing the detector system for high radiation rate applications, such as CT.

Except for switching between Geiger mode operation and proportional mode operation of the detector elements 3, the bias voltage regulatory system described above can be arranged to stabilize the gain in either mode when the radiation intensity is changing. If, for example, a protective resistor is connected in series with the bias regulator 13, a varying current will cause changes of the voltage drop across the detector elements 3 and hence in the gain. This can be compensated for by the bias regulatory system. The bias regulatory system may also be arranged to control the bias voltage $V_B$ applied to the detector elements 3 in dependence of parameters not mentioned above. For example, temperature variations of the detector elements 3 may be measured and the bias regulatory system may be arranged to vary the applied bias voltage in dependence of the measured temperature values.
Referring now to Fig. 2, a photo detector element 3 is schematically illustrated. As described above, the detector elements 3 may comprise a single APD 2 or a plurality of APDs 2, constituting a SiPM 10. According to the preferred embodiment of the present invention the detector elements 3 are SiPMs 10, as shown in Fig. 2.

The SiPM 10 comprises a matrix of APDs 2 connected in parallel. The pixel density and the active surface of each SiPM device may vary but is approximately 100-2000 or more pixels per mm², and 0.1-25 mm², respectively. When operated in Geiger mode, i.e. when the applied bias voltage $V_B$ is above the breakdown voltage of the APDs 2, one or a few secondary photons trigger an avalanche breakdown of the APD 2, resulting in an electric pulse whose magnitude is independent of the energy of the incident photons. However, the pulses from each APD 2 are added and due to the large number of pixels per unit area, the magnitude of the total pulse outputted from the SiPM device can, up to a certain number of incident secondary photons within the single-pixel recovery time, be substantially proportional to the number of secondary photons striking that particular SiPM device 10 and hence proportional to the incident primary photon energy striking the scintillators 1. Thus, the output from the SiPM device 10 can emulate the linear response of normal photomultiplier tubes and APDs operated in proportional mode.

Another advantage with the SiPM device 10 is that a signal gain similar to that of photomultiplier tubes ($10^6$-$10^7$) can be achieved with a bias of only around 20-80 volts. The high gain provided by the SiPM 10 reduces the cost of the electronics chain in the detector system. A high cost of the electronics usually limits the number of electronics channels in an imaging system, which results in reduced throughput and lower sensitivity. The SiPM device 10 is also rugged, compact (millimeters in size), and inexpensive.

As mentioned above, the bias voltage $V_B$ applied to the detector elements 3 may be chosen by an operator of the detection system and be supplied to the detector elements 3 by an ordinary voltage supply 13. But the bias voltage $V_B$ may also be automatically controlled by the bias regulator 13 in dependence of certain parameter values relating to the radiation intensity $I_t$ incident on the detector elements 3. Below, a method for measuring photon fluxes within a large dynamic range, utilizing such automatic control of the bias voltage $V_B$ is described.
In Fig. 3, a flowchart illustrating a method for measuring photon fluxes within a large dynamic range according to the present invention is shown. The method can be used in, e.g., CT, PET, SPECT, or radiotherapy portal imaging applications, or any combination of these. However, a person skilled in the art appreciates that the method is not limited to any particular application, but can be implemented in any detection system utilizing APDs or SiPM devices as detector elements.

In step 1, at least one detector element 3 is arranged to detect the radiation passing through a target region 8 emitted from an external radiation source, or radiation arising from positron annihilation or other physical processes within the target region 8. The detector element 3 may be arranged to detect this primary radiation directly or being optically coupled to for example scintillators for detecting secondary radiation that is generated by the scintillators and proportional to the primary radiation. The detector element 3 is an APD 2 or a matrix of APDs constituting a SiPM device 10.

In step 2, a controlled bias voltage $V_B$ is applied to the detector element 3.

In step 3, the magnitude and rate of the electric pulses and the mean current generated by the detector element 3 as response to the incident photon intensity $I_1$ are measured. According to the embodiment illustrated in Fig. 1, the output from the detector element 3 is registered by a data processing unit 15, preferably after being processed by preamplifiers 5, 7 and signal processing circuitries 9, 11 adapted to optimize the mean current and pulse readout.

In step 4, at least one parameter value indicative of the photon flux intensity $I_1$ that is to be measured by the detector element 3 is provided to the bias regulator 13 for regulating the bias voltage $V_B$ applied to the detector element 3. The parameter value may be provided by the radiation source 4, or by a control unit controlling the radiation source 4, and relate to, e.g., the radiation intensity $I_R$ emitted by the radiation source 4. It can also be provided by the data processing unit 15 and relate to the registered magnitude and/or rate of the electric pulses or the mean current outputted from the detector element 3.

The bias regulator 13 comprises means for comparing the received parameter value to a predetermined threshold value and increase or decrease the bias voltage $V_B$ applied to the detector element 3 in response to the result of the comparison. In Fig. 3, the comparison is
taking place in step 4 and depending on the result of the comparison, the bias voltage $V_B$ of the detector element 3 is increased, decreased (‘Yes’ in step 4) or unchanged (‘No’ in step 4). For example, the bias regulator 13 may be arranged to receive a value from the data processing unit 15 relating to the mean current outputted from the detector element 3. If the mean current is below a predetermined threshold value, indicating that the detected photon flux intensity $I_t$ is low, the bias regulator 13 applies a bias voltage $V_B$ to the detector element 3 that exceeds its breakdown voltage, thus putting the detector element 3 into Geiger mode to achieve a high internal gain. If the mean current is above the predetermined threshold value on the other hand, the bias regulator 13 applies a bias voltage $V_B$ below the breakdown voltage of the detector element 3, thus putting it into proportional mode in order to avoid non-linearities and saturation in the current readout. If the detector element 3 already operates in the operational mode that is optimal for the photon flux intensity indicated by the parameter value, no regulation of the applied bias voltage is necessary. However, optionally a SiPM gain stabilization can be achieved by fine tuning the applied bias voltage as a function of the mean current and possibly also as a function of the temperature of the detector unit.

Although some specific examples have been given above, a person skilled in the art would appreciate that there are numerous possible parameter values that may be used as indicator of the photon flux and that the invention is not limited to those examples given. Such a parameter value may be derived in many different ways and may also be provided by other devices than those shown in Fig. 1. It should also be understood that the comparison between the parameter value and any threshold value can be performed by other devices or logic circuitries than the bias regulator 13. For example, the decisions of whether or not to change the operational mode of the detector elements 3, i.e. change the bias voltage $V_B$ applied thereto, may be made by the data processing unit 15 and in turn communicated to the bias regulator 13. As aforementioned, in the most straight-forward application, the decision is simply made by the operator of the equipment and effectuated by means of switching the bias voltage between preset voltage values corresponding to Geiger mode operation and proportional mode operation, respectively.

According to one embodiment of the present invention, one single bias regulator 13 is arranged to control the bias voltage $V_B$ of all the photo detector elements 3 in the detector system. According to another embodiment, a plurality of bias regulators 13 is utilized in order to apply different bias voltages $V_B$ to different detector elements 3. By utilizing a plurality of
bias regulators 13, each connected to a single or a few detector elements 3, the bias voltage $V_B$ applied to individual detector elements 3 can be altered in dependence of local radiation intensity variations. In such a case it would be possible for a detector element 3 located in an area with a high intensity $I_I$ of incident radiation to be operated with a bias voltage $V_B$ below the breakdown voltage, thus optimizing the readout in current mode, while a detector element 3 located in an area with a low to moderate intensity $I_I$ of incident radiation may be operated with a bias voltage $V_B$ above the breakdown voltage, thereby optimizing the readout in pulse mode.

In step 5, the data processing unit 15 utilizes the optimized outputs from all the detector elements 3, representing the energy and/or timing of the detected photons and/or the mean currents induced in the photo detectors, to create diagnostic images (CT/PET/SPECT imaging) and/or therapeutic images (portal imaging) of the target region 8 in a known way. It is to be understood that the detection system and method according to the present invention may be used for any applications concerned with detecting electromagnetic radiation using APDs or SiPM devices. Although the detection system has been described in the context of a combined CT/PET/SPECT scanner for diagnostic and therapeutic imaging, it may as well be employed in, e.g., industrial radiography applications, environmental radiation monitoring, nuclear safeguard applications, airport security applications, and optical imaging applications. The radiation may as well originate from a non-controllable radiation source, such as radioactive contamination, as from a controlled or known radiation source. That is, there may not be any radiation source to “control” or “direct” or even a “target region” to examine as in the case with the combined CT/PET/SPECT application described herein. Moreover, those skilled in the art will recognize that the present invention is not limited to the exemplary configuration disclosed and illustrated herein. Different detector configurations and electronic circuitries may also be used to implement the invention. Thus, it is to be understood that the detailed disclosure of the invention only is illustrative and exemplary and merely serves the purpose of providing a full and enabling disclosure thereof. Accordingly, it is intended that the invention should be limited only by the scope of the claims appended hereinafter.
CLAIMS

1. A detection system for photon sensing and for measuring photon fluxes comprising:

   at least one detector element (3) comprising at least one avalanche photo diode [APD] (2), said detector element (3) being arranged to detect incident radiation,

   measuring means (5, 7, 9, 11, 15) for measuring the rate and magnitude of the discrete electric pulses, and the mean current, generated by the detector element (3) as a result of the incident radiation,

   characterized in that the detection system further comprises:

   a bias regulator (13) comprising means for altering a bias voltage \( V_B \) applied to the detector element (3) between at least a first and a second voltage level, said first voltage level being below the breakdown voltage of the detector element (3) for optimizing the detection system for mean current measurements and thereby for high radiation intensity applications, and said second voltage level being above the breakdown voltage of the detector element (3) for optimizing the detection system for pulse rate and magnitude measurements and thereby for low radiation intensity applications.

2. A photon detection system according to claim 1, wherein the detector element (3) comprises a plurality of APDs (2), constituting a silicon photomultiplier [SiPM] device (10).

3. A photon detection system according to claim 1 or 2, wherein the bias regulator (13) comprises an operator input device for switching between the at least two different bias voltage \( V_B \) levels.
4. A photon detection system according to any of the preceding claims, wherein the bias regulator (13) is arranged to automatically switch between the at least two different bias voltage ($V_B$) levels in dependence of a parameter value indicative of the radiation intensity ($I_t$) incident on the detector element (3).

5. A photon detection system according to claim 4, wherein the bias regulator (13) is arranged to apply a bias voltage ($V_B$) above the breakdown voltage of the detector element (3) if the parameter value is indicative of a low incident radiation intensity ($I_t$), and a bias voltage ($V_B$) below the breakdown voltage of the detector element (3) if the parameter value is indicative of a high incident radiation intensity ($I_t$).

6. A photon detection system according to any of the claims 4 or 5, wherein the bias regulator (13) is arranged to compare the at least one parameter value with a predetermined threshold value, and apply a bias voltage ($V_B$) above or below the breakdown voltage of the detector element (3) in response to the result of the comparison.

7. A photon detection system according to any of the preceding claims 4 to 6, wherein the bias regulator (13) is arranged to receive the at least one parameter value from the measuring means (7, 11, 15), said parameter value relating to the rate and/or magnitude of the discrete electric pulses generated by the detector element (3) as a result of the incident radiation.

8. A photon detection system according to any of the preceding claims 4 to 7, wherein the bias regulator (13) is arranged to receive the parameter value from the measuring means (5, 9, 15), said parameter value relating to the mean current generated by the detector element (3) as a result of the incident radiation.
9. A photon detection system according to any of the preceding claims 4 to 8, wherein the incident radiation is primary or secondary radiation from a controlled radiation source (4), the bias regulator (13) being arranged to receive the parameter value from the radiation source (4) or a radiation source control unit, said parameter value relating to the radiation intensity $I_R$ emitted from said radiation source (4).

10. A photon detection system according to any of the preceding claims, wherein the measuring means (5, 7, 9, 11, 15) comprises a data processing unit (15) arranged to receive and register the pulse signals and the mean current signal generated by the detector element (3) as a result of the incident radiation.

11. A photon detection system according to any of the preceding claims, wherein the measuring means (5, 7, 9, 11, 15) comprises at least one preamplifier (5, 7) arranged to amplify the pulse signals and/or the mean current signal generated by the detector element (3) as a result of the incident radiation, before being received by the data processing unit (15).

12. A photon detection system according to any of the preceding claims, wherein the measuring means (5, 7, 9, 11, 15) comprises a pulse processing circuitry (11) arranged to optimize the pulse signals, and a current processing circuitry (9) arranged to optimize the mean current signal, generated by the detector element (3) as a result of the incident radiation, before being received by the data processing circuitry (15).

13. A photon detection system according to any of the preceding claims, comprising a plurality of detector elements (3), all being operated with a bias voltage ($V_B$) controlled by a common bias regulator (13).
14. A photon detection system according to any of the claims 1 to 12, wherein the detection system comprises a plurality of detector elements (3) and a plurality of bias regulators (13), all bias regulators (13) controlling a bias voltage \( V_B \) applied to different detector elements (3) or different groups of detector elements (3).

15. A photon detection system according to any of the claims 13 or 14, wherein the pulse output and/or the current output from several detector elements (3) are added before being measured by the measuring means (5, 7, 9, 11, 15).

16. A photon detection system according to claim 15, wherein the pulse output and/or the current output from each detector element (3) is amplified by the at least one preamplifier (5, 7) before being added.

17. A photon detection system according to any of the preceding claims, wherein the detector element (3) is optically coupled to at least one scintillator (1), the detector element (3) being arranged to detect secondary radiation generated by said at least one scintillator (1) as response to primary radiation incident on the scintillator (1).

18. A photon detection system according to any of the preceding claims, wherein a plurality of detector elements (3) are optically coupled to one scintillator (1), the detector elements (3) being arranged to detect secondary radiation generated by said at least one scintillator (1) as response to primary radiation incident on the scintillator (1).

19. A photon detection system according to any of the preceding claims, wherein the bias regulator (13) is arranged to receive a first parameter value from the measuring means (5, 9, 15) relating to the mean current generated by the detector element (3) as a result of the incident radiation, and/or to receive a second
parameter value from a temperature sensor relating to the temperature of the
detector element (3), said bias regulator (13) further being arranged to
automatically fine tune the bias voltage ($V_b$) applied to the detector element (3)
around the at least first or second voltage level based on said first and/or second
parameter value.

20. A scanner for computed tomography [CT], positron emission tomography [PET],
single photon emission computed tomography [SPECT], or radiotherapy portal
imaging applications, or any combination of these, characterized in that it
comprises a photon detection system according to any of the claims 1 to 19.

21. Use of a photon detection system according to any of the claims 1 to 19 in positron
emission tomography [PET], photon emission computed tomography [SPECT],
computed tomography [CT], or radiotherapy portal imaging applications, or any
combination of these.

22. A method for photon sensing and for measuring photon fluxes comprising the
following steps:

     arranging at least one detector element (3) comprising at least one avalanche
     photo diode [APD] (2) to detect incident radiation;

     measuring the rate and magnitude of the discrete electric pulses, and the mean
current, generated by the at least one detector element (3) as a result of the detected
radiation,

     characterized in that the method further comprises the step of:

     altering a bias voltage ($V_b$) applied to the detector element (3) between at least a
first and a second voltage level in dependence of the radiation intensity ($I_0$) incident
on the detector element (3), said first voltage level being below the breakdown
voltage of the detector element (3) for optimizing the mean current measurements
and thereby optimizing the method for high radiation intensity applications, and said second voltage level being above the breakdown voltage of the detector element (3) for optimizing the pulse rate and magnitude measurements and thereby optimize the method for low radiation intensity applications.

23. Method according to claim 22, wherein the step of altering the bias voltage ($V_B$) applied to the detector element (3) is performed by manually regulating an operator input device of a bias regulator (13).

24. Method according to claim 22 or 23, wherein the step of altering the bias voltage ($V_B$) applied to the detector element (3) is automatically performed by a bias regulator (13) in dependence of a parameter value indicative of the radiation intensity ($I_i$) incident on the detector element (3).

25. Method according to claim 24, wherein the step of altering the bias voltage ($V_B$) further comprises the step of:

- applying a bias voltage ($V_B$) above the breakdown voltage of the detector element (3) if the parameter value is indicative of a low incident radiation intensity ($I_i$), and a bias voltage ($V_B$) below the breakdown voltage of the detector element (3) if the parameter value is indicative of a high incident radiation intensity ($I_i$).

26. Method according to claim 24 or 25, wherein the step of altering the bias voltage ($V_B$) further comprises the steps of:

- comparing the at least one parameter value with a predetermined threshold value, and

- applying a bias voltage ($V_B$) above or below the breakdown voltage of the detector element (3) in response to the result of the comparison.
27. Method according to any of the claims 24 to 26, wherein the at least one parameter value relates to the rate and/or magnitude of the discrete electric pulses generated by the detector element (3) as a result of the incident radiation.

28. Method according to any of the claims 24 to 27, wherein the at least one parameter value relates to the mean current generated by the detector element (3) as a result of the incident radiation.

29. Method according to any of the claims 24 to 28, wherein the radiation incident on the detector element (3) is primary or secondary radiation from a controlled radiation source (4), the at least one parameter value relating to the radiation intensity \( I_R \) emitted from said radiation source (4).

30. Method according to any of the claims 22 to 29, wherein the step of measuring the rate and magnitude of the discrete electric pulses, and the mean current, outputted from the detector element (3) further comprises the steps of:

   amplifying the pulse signals and/or the mean current signal by means of at least one preamplifier (5, 7),

   optimizing the pulse signals and the mean current signal by means of a pulse processing circuitry (11) and a current processing circuitry (9), respectively, and

   register the pulse signals and the mean current signal outputted from the pulse processing circuitry (11) and the current processing circuitry (9) by means of a data processing unit (15).
31. Method according to any of the claims 22 to 30, wherein a plurality of detector elements (3) are arranged to detect incident radiation, all detector elements (3) being operated with a common controlled bias voltage (V_B).

32. Method according to any of the claims 22 to 30, wherein a plurality of detector elements (3) are arranged to detect incident radiation, the bias voltage (V_B) of each detector element (3) or some detector elements (3) being separately controlled in order to apply different bias voltages (V_B) to detector elements (3) exposed to different radiation intensities (I_i).

33. Method according to claim 31 or 32, wherein the pulse output and the mean current output, respectively, from several detector elements (3) are added before being measured.

34. Method according to any of the claims 22 to 33, wherein the step of arranging the at least one detector element (3) to detect incident radiation comprises the step of:

   arranging the detector element (3) in optical contact with at least one scintillator (1) to detect secondary radiation generated by the least one scintillator (1) as response to primary radiation incident on said at least one scintillator (1).

35. Method according to any of the claims 22 to 34, further comprising the step of:

   fine tuning the bias voltage (V_B) applied to the detector element (3) around the at least first or second voltage level in dependence of a first parameter value relating to the mean current generated by the detector element (3) as a result of the incident radiation and/or a second parameter value relating to the temperature of the detector element (3).
FIG 3

1. Arrange APD/SiPM detector element(s) to detect primary or secondary incident radiation.

2. Apply a controlled bias voltage to the APD/SiPM detector element(s).

3. Measure the magnitude and rate of the electric pulses and the mean current generated by the APD/SiPM detector element(s) as response to the incident radiation.

4. Increase/decrease the bias voltage applied to the APD/SiPM detector element(s) in order to change operational mode thereof.

   - Yes: Parameter value being indicative of a photon flux requiring the opposite operational mode of the APD/SiPM detector element(s) to optimize readout?
   - No: Go back to step 3.

5. Use pulse and/or current data to create diagnostic and/or therapeutic images of the target region.
### INTERNATIONAL SEARCH REPORT

**International application No.**

PCT/SE2008/050373

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**A. CLASSIFICATION OF SUBJECT MATTER**

**IPC:** see extra sheet

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

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**B. FIELDS SEARCHED**

Minimum documentation searched (classification system followed by classification symbols)

**IPC:** G01J, H01L, A61B

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 practical, search terms used)

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**C. DOCUMENTS CONSIDERED TO BE RELEVANT**

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**Date of the actual completion of the International search** 14 July 2008

**Date of mailing of the international search report** 17-07-2008

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International patent classification (IPC)
G01J 1/44 (2006.01)
A61B 6/03 (2006.01)
H01L 31/107 (2006.01)

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