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(71) **Applicant** *(for all designated States except US)*: **KONINKLIJKE PHILIPS ELECTRONICS, N.V.** [NL/NL]; Groenewoudseweg 1, NL-5621 BA Eindhoven (NL).

(72) **Inventors; and**

(75) **Inventors/Applicants** *(for US only)*: **LUHTA, Randall, P.** [CA/US]; 5700 Highland Road, Highland Heights, OH 44143 (US). **CHAPPO, Marc, A.** [US/US]; 1057 Bell Avenue, Elyria, OH 44035 (US). **HARWOOD, Brian, E.** [US/US]; 1704 Mount Carmel Drive, Wickliffe, OH 44094 (US). **MATTSON, Rodney, A.** [US/US]; 6532 Melshore Drive, Mentor, OH 44060 (US). **VRETTOS, Chris, J.** [US/US]; 5739 Noble Court, Willoughby, OH 44094 (US).

(74) **Common Representative: KONINKLIJKE PHILIPS ELECTRONICS, N.V.**; c/o LUNDIN, Thomas, M., 595 Miner Road, Cleveland, OH 44143 (US).

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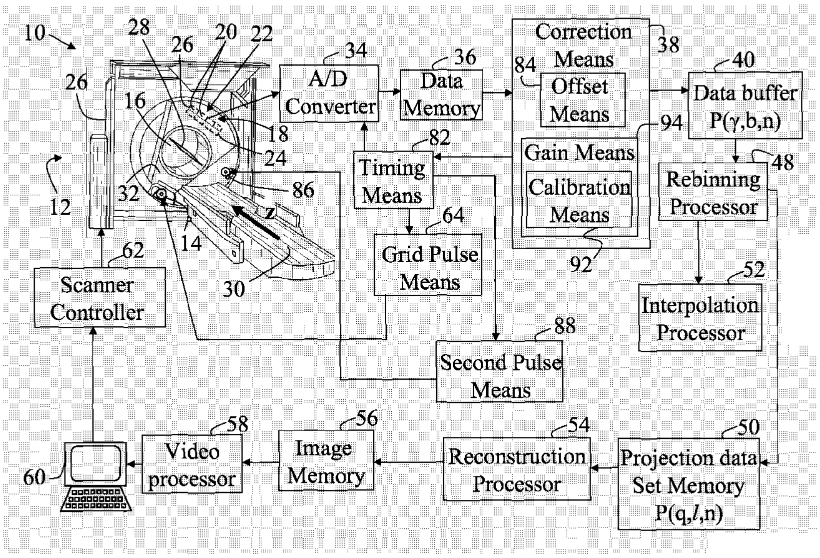
(54) **Title:** PULSED X-RAY FOR CONTINUOUS DETECTOR CORRECTION



(57) **Abstract:** A radiographic imaging apparatus (10) comprises a primary radiation source (14) which projects a beam of radiation into an examination region (16). A detector (18) converts detected radiation passing through the examination region (16) into electrical detector signals representative of the detected radiation. The detector (18) has at least one temporally changing characteristic such as an offset B(t) or gain A(t). A grid pulse means (64) turns the primary radiation source (14) ON and OFF at a rate between 1000 and 5000 pulses per second, such that at least the offset B(t) is re-measured between 1000 and 5000 times per second and corrected a plurality of times during generation of the detector signals. The gain A(t) is measured by pulsing a second pulsed source (86, 100, 138) of a constant intensity (XRef) with a second pulse means (88). The gain A(t) is re-measured and corrected a plurality of times per second during generation of the detector signals.

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PULSED X-RAY FOR CONTINUOUS DETECTOR CORRECTION

DESCRIPTION

The present application relates to the diagnostic imaging arts. It finds particular application in the computed tomography imaging, and will be described with particular reference thereto. However, it also finds application in other imaging apparatuses and methods that employ x-ray detectors.

CT scanners typically include an x-ray source and arrays of x-ray detectors secured respectively on diametrically opposite sides of a gantry. During a scan of a patient located within the bore of the gantry, the gantry rotates about a rotation axis while an x-ray source emits x-rays during the data collection period of the scan. The x-rays are collected by the detector which includes a plurality of detector elements.

Typically, the x-ray detector used in a CT scanner includes a layer of scintillating crystals which is coupled to an array of silicon photodiodes. The scintillating crystals absorb x-rays that have passed through the patient and produce light in proportion to the intensity of the absorbed x-rays. The photodiodes absorb the light produced by the scintillating crystals and convert it into an electrical current in proportion to the light absorbed. The ideal detector produces a signal current S(t) in direct proportion to the x-ray intensity X incident to the detector.

Generally, x-ray detectors are characterized by a time dependent gain **A(t)** and offset **B(t)**. Offset is represented by a residual signal which stays ON after the x-ray source is turned OFF. Typically, the scintillating layer of x-ray detector is selected from the materials which have the gain **A(t)** and offset **B(t)** nearly constant with time. Most often, for such a detector (with nearly constant gain **A(t)** and offset **B(t)**), no correction for changes in gain and/or offset during a scan are made. In some scanning systems, only a minor correction for changes in offset **B(t)** is made during a scan. Usually, the offset **B(t)** of the detector is calibrated once every scan, e.g. approximately every 30 seconds. The gain **A(t)** of the detector may be calibrated once every month. Therefore, the materials for the detectors are carefully selected such that the changes in offset and gain are so minor between calibrations than can be neglected.

Scintillating crystal materials, for which gain and offset are nearly constant, are expensive which adds a substantial cost to the cost of the detector. However, if the lower grade, lower cost scintillating crystal materials are used in a CT scanner, the changes in gain **A(t)** and offset **B(t)** during a scan may be substantial and cannot be neglected.

Another problem encountered in the modern CT scanners is the reduction in signal per detector as the detectors are made smaller with correspondingly thinner slices. The reduction in signal per detector can cause artifacts in CT scans where high attenuation of the patient results in low signal conditions, e.g. when the patient is imaged through the shoulders. The low signal condition can be overcome by using a detector with a higher gain $\mathbf{A(t)_x}$, thus improving the signal-to-noise ratio.

The materials with a high gain are known. Such materials could have been used in the CT detector, if it were not for the known problem of the change in offset and/or gain. For example, the direct conversion semiconductors (x-ray photoconductors), such as CdZnTe, CdTe, TlBr, PbO and the like, can provide a gain as much as ten times greater than the scintillator-photodiode detectors. The higher gain could improve imaging in the scanners with thinner slices. However, the gain and offset of the photoconductors are not stable and vary substantially with time.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect of the present application, a radiographic imaging apparatus is disclosed. A primary radiation source projects a beam of radiation into an examination region in which a subject is disposed for an examination. A detector converts detected radiation passing through the examination region into electrical detector signals representative of the detected radiation and at least one temporally changing characteristic. A correction means determines a correction to the detector signals to compensate for the at least one temporally changing characteristic a plurality of times during generation of the detector signals and corrects the detector signals with the determined corrections.

According to another aspect, a radiographic imaging method is disclosed. A beam of radiation is projected into an examination region in which a subject is disposed for an examination. Detected radiation passing through the examination region is converted into electrical detector signals representative of the detected radiation and at least one temporally changing characteristic. A correction to the detector signals is determined to compensate for the at least one temporally changing characteristic a plurality of times during generation of the detector signals. The detector signals are corrected with the determined corrections.

One advantage of the present application resides in continuously correcting the gain and offset of an x-ray detector during the CT scan.

Another advantage resides in utilizing existing imaging detectors.

Another advantage resides in utilizing existing conversion electronics.

Another advantage resides in not adding time to the scan.

Another advantage resides in effectively preventing artifacts in the image due to a change in offset and/or gain of detector during one scan.

Another advantage resides in utilizing lower cost detector materials which are known to have time varying characteristics.

Another advantage resides in reducing a thickness of the scanner slice by using x-ray photoconductors.

Numerous additional advantages and benefits will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiments.

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 shows a diagrammatic representation of a computed tomography imaging system;

FIGURE 2 shows a diagrammatic representation of a portion of an imaging system which includes a pulsed x-ray source;

FIGURE 3A shows a timing diagram of data collection;

FIGURE 3B shows a timing diagram of pulsing x-ray source;

FIGURE 4 diagrammatically shows a portion of the CT system with a pulsed primary x-ray source and a pulsed secondary x-ray source;

FIGURE 5A shows a timing diagram of a data collection for a gain calibration;

FIGURE 5B shows a timing diagram of pulsing the secondary source during a gain calibration;

FIGURE 5C shows a timing diagram of data collection during scan for the imaging system which includes two pulsed sources;

FIGURE 5D shows a timing diagram of pulsing a primary x-ray source during scan;

FIGURE 5E shows a timing diagram of pulsing a secondary source during scan;

FIGURE 6A diagrammatically shows a field of view of a detector;

FIGURE 6B shows a graph of average x-ray intensity which is modulated by a fast pulsing of an x-ray source;

FIGURE 7 diagrammatically shows a detailed portion of the CT imaging system in which the secondary pulsed source is a light source;

FIGURE 8 diagrammatically shows a detailed portion of the CT imaging system in which the detector includes an x-ray photoconductor and the secondary pulsed source is a light source; and

FIGURE 9 diagrammatically shows a detailed portion of the CT imaging system in which the detector includes an x-ray photoconductor and the secondary pulsed source is a charge carrier injection source.

With reference to FIGURE 1, an imaging system **10** includes a computed tomography scanner **12** which houses or supports a primary radiation source **14**, which in one embodiment, is an x-ray source or x-ray tube, that projects a radiation beam into an examination region **16** defined by the scanner **12**. After passing through the examination region **16**, the radiation beam is detected by a two-dimensional radiation detector **18** which includes a plurality of detection modules or detection elements **20** arranged to detect the radiation beam after passing through the examination region **16**. The detector **18** includes an x-ray to analog signal converting layer **22** which has a gain **A(t)** and/or offset **B(t)** characteristics that typically change with time. In one embodiment, the converting layer **22** includes an array of scintillating crystals or a scintillator or a scintillating layer **24** which is coupled with an array of photodiodes **26**. In another embodiment, the converting layer **22** includes a plurality of direct conversion semiconductors or x-ray photoconductors such as CZT, CdTe, TlBr, PbO and the like. Typically, the x-ray tube **14** produces a diverging x-ray beam having a cone beam, wedge beam, or other beam geometry that expands as it passes through the examination region **16** to substantially fill the area of the radiation detector **18**.

An imaging subject **28** (shown in the examination region **16**) is placed on a couch **30** or other support that moves the imaging subject into the examination region **16**. The couch **30** is linearly movable along an axial direction $\mathbf{O_Z}$ (designated as a **Z**-direction in FIGURE 1.) The radiation source **14** and the radiation detector **18** are oppositely mounted with respect to the examination region **16** on a rotating gantry **32**, such that rotation of the gantry **32** effects revolving of the radiation source **14** about the examination region **16** to provide an angular range of views. The acquired data is referred to as projection data since each detector element detects a signal corresponding to an attenuation line integral taken along a line, narrow cone, or other substantially linear projection extending from the source to the detector element.

In one embodiment, an axial projection data set is acquired with the rotating gantry **32** rotating while the couch **30** is stationary. The axial projection data set includes a plurality of axial slices corresponding to rows or columns of detector elements transverse to the axial or **Z**-direction. Optionally, additional axial slices are acquired by performing repeated axial scans and moving the couch **30** between each axial scan.

In another embodiment, a helical projection data set is acquired by rotating the gantry **32** simultaneously with continuous linear motion of the couch **30** to produce a helical trajectory of the radiation source **14** around the imaging subject disposed on the couch **30**.

During scanning, some portion of the radiation passing along each projection is absorbed by the imaging subject to produce a generally spatially varying attenuation of the radiation. The detection elements **20** of the detector **18** sample the radiation intensities across the radiation beam to generate radiation absorption projection data. An analog to digital converter **34** converts the analog signal collected by the detector **18** into a series of digital numbers. The digital data including both x-ray attenuation measurements of the subject, offset measurements, and gain measurements is stored in a data memory **36**. A correction means **38** applies a mathematical correction to the projection data to correct for changes in the detector offset **B(t)** and/or gain **A(t)** which vary with time, as will be discussed in a greater detail below. In one embodiment, the correction means **38** is incorporated with the analog to digital converter **34**. The corrected projection data is stored in a buffer memory **40**.

For a source-focused acquisition geometry in a multi-slice scanner, readings of the attenuation line integrals or projections of the projection data set stored in the buffer memory **40** can be parameterized as $P(\gamma,\beta,n)$, where γ is the source angle of the radiation source **14** determined by the position of the rotating gantry **32**, β is the angle within the fan ($\beta \in [-\Phi/2, \Phi/2]$, where Φ is the fan angle), and n is the detector row number in the O_Z direction. In one embodiment, a rebinning processor **48** rebins the projection data into a parallel, non-equidistant raster of canonic trans-axial coordinates. The rebinning can be expressed as $P(\gamma,\beta,n) \rightarrow P(\theta,l,n)$, where θ parameterizes the projection number that is composed of parallel readings parameterized by l which specifies the distance between a reading and the isocenter, and n is the detector row number in the O_Z direction.

The rebinned parallel ray projection data set $P(\theta,l,n)$ is stored in a projection data set memory **50**. Optionally, the projection data is interpolated by a interpolation processor **52** into equidistant coordinates or into other desired coordinates spacings before storing the projection data $P(\theta,l,n)$ in the projection data set memory **50**. A reconstruction processor **54** applies filtered backprojection or another image reconstruction technique to reconstruct the projection data set into one or more reconstructed images that are stored in a reconstructed image memory **56**. The reconstructed images are processed by a video processor **58** and displayed on a user interface **60** or is otherwise processed or utilized. In one embodiment, the user interface **60** also enables a radiologist, technician, or other operator to interface with a computed tomography scanner controller **62** to implement a selected axial, helical, or other computed tomography imaging session.

With continuing reference to FIGURE 1 and further reference to FIGURE 2, a grid pulse circuit or means **64** turns the x-ray tube **14** ON and OFF at a rate of about 1000 to 5000 pulses per second to allow a time varying offset **B(t)** to be measured the same number of times per second, i.e., continuously over time. At a sampling rate of 1000-5000 pulses per second, the offset **B(t)** only needs to be considered constant for a time on the order of 100's of microseconds. This assumption is quite reasonable for a wide range of the low grade scintillators and x-ray photoconductors that might be used with to manufacture the detector **18**. Preferably, a pulse rise time $\mathbf{T_r}$ and a pulse fall time $\mathbf{T_f}$ is equal to or less than 1usec.

More specifically, the fast switching is achieved by quickly switching the potential on a cathode cup or grid **66** relative to a filament **68**. Turning to a circuit **A**, a positive high voltage power supply **70** is connected to an anode **72** of the x-ray tube **14**. A negative high voltage power supply **74** is connected to the cathode **66** of the x-ray tube **14**. A voltage for each of the power supplies **70, 74** is preferably about +60,000 volts/-60,000 volts, respectively, which results in a total potential from the anode **72** to the cathode **66** of 120,000 volts. The grid pulse means **64** includes a grid power supply **76**, a switch **78** and a filament power supply **80** which supply a pulse potential to the grid **66** and the filament **68**. The grid pulse voltage is preferably from 1000 to 5000 volts. As the switch **78** receives command signals from a timing control unit or means **82** to close or open, the grid pulse means **64** changes the potential on the cathode **66** relative to the potential on the filament **68**. When the voltage on the cathode **66** is made negative relative to the filament by approximately a few thousand volts, the electron beam becomes pinched off and the x-rays are distinguished. The correction means **38** measures the signal, calculates correction and applies it to the measurement.

With continuing reference to FIGURES 1 and 2 and further reference to FIGURES 3A and 3B, during an ON time $\mathbf{t_1}$ of the x-ray source **14**, the measured signal $\mathbf{S(t_1)}$ is generally equal to:

$$S(t_1) = A(t_1)\bullet X(t_1) + B(t_1), \text{ where}$$

$\mathbf{S(t_1)}$ is a measured signal of the detector **18** during the x-ray source ON time $\mathbf{t_1}$,
$\mathbf{X(t_1)}$ is an incident x-ray intensity on the detector **18**,
$\mathbf{A(t_1)}$ is a gain of the detector **18** during the ON time $\mathbf{t_1}$, and
$\mathbf{B(t_1)}$ is a signal offset of the detector **18** during the ON time $\mathbf{t_1}$.

During a time $\mathbf{t_2}$, the x-ray source **14** is OFF, an offset measuring means **84** measures the offset $\mathbf{B(t_2)}$ which is equal to the measured signal $\mathbf{S(t_2)}$:

$$S(t_2) = B(t_2), \text{ where}$$

$S(t_2)$ is a measured signal of the detector **18** during the OFF time **t_2**, and
$B(t_2)$ is a measured offset of the detector **18** during the OFF time **t_2**.

Because the ON time **t_1** and the OFF time **t_2** are close in time to each other, e.g. within 200 usec, the change in offset is only minor and can be neglected. E.g., the value of the measured offset **$B(t_2)$** is nearly equal to the value of the signal offset **$B(t_1)$**:

$$B(t_2) \approx B(t_1)$$

The correction means **38** calculates a corrected signal value by subtracting the measured offset **$B(t_2)$** from the measured signal **$S(t_1)$** to obtain the corrected signal **$S_C(t_1)$**:

$$S_C(t_1) = [A(t_1) \bullet X(t_1) + B(t_1)] - B(t_2) \quad \approx A(t_1) \bullet X(t_1)$$

or

$$S_C(t_1) = S(t_1) - S(t_2), \text{ where}$$

$S_C(t_1)$ is a corrected signal value,
$X(t_1)$ is an incident x-ray intensity on the detector **18,**
$A(t_1)$ is a gain of the detector **18** during the ON time **t_1,**
$B(t_1)$ is a signal offset of the detector **18** during the ON time **t_1,**
$B(t_2)$ is a measured offset of the detector **18** during the OFF time **t_2,**
$S(t_1)$ is a measured signal of the detector **18** during the ON time **t_1**, and
$S(t_2)$ is a measured signal of the detector **18** during the OFF time **t_2**.

In timing diagrams of FIGURES 3A and 3B, the ON time **T_{ON}** and the OFF time **T_{OFF}** for the x-ray pulse is shown to be the same width as the data collection intervals. Of course, it is also contemplated that at least one of the ON time **T_{ON}** and OFF time **T_{OFF}** for the x-ray pulse can be shorter than the data collection interval, for example, to allow time for the detector signal to decay before an offset measurement is taken. Alternatively, the ON and OFF times of the x-ray source might be of different lengths of time. The respective data collection intervals for the ON and OFF times of the radiation source might vary in lengths as appropriate.

With continuing reference to FIGURE 1 and further reference to FIGURE 4, a secondary illumination source **86** such as an x-ray source, is added to the circuit **A** of FIGURE 2 to continuously measure the gain **A(t)** of the detector **18** with the same timing as the offset measurements. A measurement of the gain can be done by turning OFF the primary x-ray source **14** and turning ON the secondary x-ray source **86** that illuminates the detectors directly rather than through the subject. The illumination source **86** produces the secondary illumination of a constant intensity $\mathbf{X_{Ref}}$ which is used as the reference illumination to measure the detector gain **A(t)**. Preferably, the secondary x-ray source **86** is an x-ray source of a lower radiation intensity than the primary x-ray source **14**. The secondary source **86** is preferably disposed in a closer proximity to the detector **18** than the primary x-ray source **14** such that the illumination from the secondary source **86** does not pass through the subject **28**. In one embodiment, for example, the primary x-ray tube **14** is rated between 100 and 500mA, while the secondary x-ray source **86** can be rated between 1 and 2mA. Such secondary x-ray source **86** has a small footprint and is cost effective. One possible position to dispose the secondary x-ray source **86** is behind the detector **18** such that the x-rays pass through the back of the detector **18**. The secondary x-ray source **86** might be a field emission x-ray tube or fast shuttered radioisotope source.

The secondary source **86** is a grid controlled x-ray tube, similar to the primary x-ray tube **14**. A secondary pulse means **88** turns the secondary x-ray source **86** ON and OFF at a rate of about 1000 to 5000 pulses per second. Preferably, a pulse rise time $\mathbf{T'_r}$ and a pulse fall time $\mathbf{T'_f}$ are equal to or less than 1usec. An appropriate electronics **90** is included with the circuitry. The electronics **90** preferably includes a positive high voltage power supply to supply a high power to an anode of the secondary x-ray source **86**, a negative high voltage power supply to supply a negative high power to a grid of the x-ray source **86**, a grid power supply, a switch and a filament power supply to supply a pulse potential to the grid and a filament of the secondary x-ray source **86**. The secondary pulse means **88** receives command signals from the timing control unit or means **82** to close or open the switch of the secondary pulse means to change the potential to the grid relative to the filament to quickly extinguish x-rays.

With continuing reference to FIGURES 1 and 4 and further reference to FIGURES 5A and 5B, sometime prior to scanning, a calibration process or means **92** measures a nominal or calibration gain $\mathbf{A_{cal}}$ of the detector **18**. The future measurements are corrected to the nominal gain. The calibration process **92** includes two measurements that are made in quick succession. The first measurement $\mathbf{S(t_A)}$ is taken when the secondary x-ray source **86** is ON during a time $\mathbf{t_A}$. The second measurement $\mathbf{S(t_B)}$ is taken when the secondary x-ray source **86** is OFF at a time $\mathbf{t_B}$. The two measured signals $\mathbf{S(t_A)}$, $\mathbf{S(t_B)}$ have the relations:

$$S(t_A) = A_{Cal} \bullet X_{Ref} + B(t_A) \qquad (1)$$
$$S(t_B) = B(t_B), \text{ where}$$

$\mathbf{S(t_A)}$ is a measured signal of the detector during the time $\mathbf{t_A}$,
$\mathbf{X_{REF}}$ is an intensity of the secondary source **86**,
$\mathbf{A_{Cal}}$ is a calibration gain of detector **18** during the time $\mathbf{t_A}$,
$\mathbf{B(t_A)}$ is a signal offset of the detector **18** during the time $\mathbf{t_A}$,
$\mathbf{S(t_B)}$ is a measured signal of the detector **18** during the time $\mathbf{t_B}$, and
$\mathbf{B(t_B)}$ is a measured offset of the detector **18** during the time $\mathbf{t_B}$.

Because times $\mathbf{t_A}$ and $\mathbf{t_B}$ are close in time, the change in offset is only minor and can be neglected. Therefore, the measured offset $\mathbf{B(t_B)}$ is approximately equal to the signal offset $\mathbf{B(t_A)}$. The value of the calibration gain $\mathbf{A_{Cal}}$ is equal to:

$$A_{Cal} = [S(t_A) - B(t_A)]/ X_{Ref} \quad \text{or}$$
$$A_{Cal} = [S(t_A) - S(t_B)]/ X_{Ref}$$

$\mathbf{A_{Cal}}$ is a calibration gain of detector **18** during the time $\mathbf{t_A}$,
$\mathbf{S(t_A)}$ is a measured signal of the detector during the time $\mathbf{t_A}$,
$\mathbf{B(t_A)}$ is a signal offset of the detector **18** during the time $\mathbf{t_A}$,
$\mathbf{X_{REF}}$ is an intensity of the secondary source **86**, and

$S(t_B)$ is a measured signal of the detector **18** during the time **t_B**.

With continuing reference to FIGURES 1 and 4 and further reference to FIGURES 5C-5E, a gain measuring means **94** calculates the gain of the detector **18**. Initially, the secondary x-ray source **86** is OFF. During the time **t_1**, the primary x-ray source **14** is ON. The measured signal **$S(t_1)$** during the time **t_1** is equal to:

$$S(t_1) = A(t_1) \bullet X(t_1) + B(t_1), \text{ where}$$

$S(t_1)$ is a measured signal of the detector **18** during the time **t_1**,
$X(t_1)$ is an incident x-ray intensity on the detector **18**,
$A(t_1)$ is a gain of the detector **18** during the time **t_1**, and
$B(t_1)$ is a signal offset of the detector **18** during the time **t_1**.

As explained above, during the OFF time **t_2**, the offset measuring means **84** measures the offset **$B(t_2)$**, which is equal to the measured signal **$S(t_2)$**:

$$S(t_2) = B(t_2), \text{ where}$$

$S(t_2)$ is a measured signal of the detector **18** during the OFF time **t_2**, and
$B(t_2)$ is a measured offset of the detector **18** during the OFF time **t_2**.

During a time **t_3**, which is close in time to the times **t_1** and **t_2**, the gain measuring means **94** measures the gain of the detector **18**. More specifically, the timing means **82** sends a command to turn OFF the x-ray source **14** and turn ON the secondary x-ray source **86**. The measured signal **$S(t_3)$** during the time **t_3** is equal to:

$$S(t_3) = A(t_3) \bullet X_{Ref} + B(t_3), \text{ where}$$

$S(t_3)$ is a measured signal of the detector **18** during the time **t_3**,
$A(t_3)$ is a signal gain of the detector **18** during the time **t_3**,
X_{REF} is an intensity of the secondary source **86**, and
$B(t_3)$ is a signal offset of the detector **18** during the time **t_3**.

The corrected signal **$S_C(t_1)$**, which is a close approximation to the desired signal is equal to:

$$S_C(t_1) = A_{Cal} \bullet X(t_1), \text{ where}$$

$S_C(t_1)$ is a corrected signal of the detector during the time **t_1**,
$X(t_1)$ is an incident x-ray intensity on the detector **18**, and
A_{Cal} is a calibration gain of detector **18**.

If the calibration gain **A_{cal}** is equal to $[S(t_A) - S(t_B)]/ X_{Ref}$ and the x-ray intensity **$X(t_1)$** is equal to $[S(t_1) - B(t_1)]/A(t_1)$, then the corrected signal **$S_C(t_1)$** is equal to:

$$S_C(t_1) = \{[S(t_A) - S(t_B)]/ X_{Ref}\} \bullet \{[S(t_1) - B(t_1)]/ A(t_1)\}, \text{ where}$$

$S_C(t_1)$ is a corrected signal of the detector **18** during the time **t_1**,
$S(t_A)$ is a measured signal of the detector **18** during the time **t_A**,
$S(t_B)$ is a measured signal of the detector **18** during the time **t_B**,
X_{REF} is an intensity of the secondary source **86**,
$S(t_1)$ is a measured signal of the detector **18** during the time **t_1**,
$B(t_1)$ is a signal offset of the detector **18** during the time **t_1**, and
$A(t_1)$ is a signal gain of the detector **18** during the time **t_1**.

An assumption here is made that the times **t_1**, **t_2** and **t_3** are close in time; e.g. the detector offset stays constant at the times **t_1, t_2, t_3**, therefore, **$B(t_1) \approx B(t_2) = S(t_2)$**, thus the corrected signal **$S_C(t_1)$** is equal to:

$$S_C(t_1) = \{[S(t_A) - S(t_B)/ X_{Ref}\} \bullet \{[S(t_1) - S(t_2)/ A(t_1)\}$$

Since $A(t_1) \approx A(t_3) = \{[S(t_3) - B(t_3)]/ XRef\}$ and $B(t_3) \approx B(t_2) = S(t_2)$, then:

$$S_C(t_1) = \{[S(t_A) - S(t_B)] \bullet [S(t_1) - S(t_2)]\} / \{S(t_3) - S(t_2)\}, \text{ where}$$

$S_C(t_1)$ is a corrected signal of the detector **18** during the time **t_1**,

$S(t_A)$ is a measured signal of the detector **18** during the time **t_A**,

$S(t_B)$ is a measured signal of the detector **18** during the time **t_B**,

$S(t_1)$ is a measured signal of the detector **18** during the time **t_1**,

$S(t_2)$ is a measured signal of the detector **18** during the time **t_2**,

$S(t_3)$ is a measured signal of the detector **18** during the time **t_3**.

Therefore, the correction means **38** corrects the gain and offset of the signal **$S(t_1)$** at the time **t_1** to yield the corrected signal **$S_C(t_1)$** which is the corrected signal at the time **t_1**. The corrected signal **$S_C(t_1)$** is a close approximation of the measured signal **$S'(t_1)$** would occur if the gain **$A(t)$** were a constant value of **A_{Ref}** and the offset was equal to zero.

In one embodiment, to reduce noise it is also possible to smooth or average the offset and gain measurements over time.

As shown in FIGURES 5C, 5D and 5E, in a timing diagram for the system which includes two pulsed x-ray sources for continuous gain **$A(t)$** and offset **$B(t)$** correction, the ON time **T_{ON}** and the OFF time **T_{OFF}** for the x-ray pulse of the x-ray source **14** is shown to be the same width as the data collection intervals. Of course, it is also contemplated that at least one of the ON time **T_{ON}** and OFF time **T_{OFF}** for the x-ray pulse of the x-ray source **14** can be shorter than the data collection interval, for example, to allow time for the detector signal to decay before an offset measurement is taken. Alternatively, the ON and OFF times of the x-ray source might be of different lengths of time. The respective data collection intervals for the ON and OFF times of the x-ray source might vary in lengths as appropriate.

With reference to FIGURES 6A and 6B, the x-ray tube **14** intensity is varied continuously to achieve a fast dose modulation and reduce the overall radiation dose delivered to the patient. In the CT scanners, it is common to vary the current to the filament **68** of the x-ray tube **14** in order to vary the x-ray intensity during a scan. More x-rays are typically required when the x-rays must pass through the thick anatomy of the subject **28**; while fewer x-rays are used when the imaging is done in the regions with a thin anatomy. When the average x-ray intensity is required to be large, e.g it is known that the x-rays have to go through a wide part of the subject **28**, the long x-ray pulses **96** are used. When less average intensity is required, e.g it is known that the x-rays have to go through a thinner part of the subject **28**, the shorter pulses **98** are used. By reducing the x-ray intensity when the x-rays pass through thin sections of the patient anatomy, the overall patient's dose is reduced. The average x-ray intensity stays nearly constant over scan.

With reference to FIGURE 7, a pulsed light source **100** such as an ultraviolet light (UV) is used to excite the scintillator **24** to introduce light into the detector **18** that is detected by the photodiodes **26**. The light injected into the scintillating layer **24** is absorbed similar to the x-rays which are absorbed in the same material. An ultraviolet light **102** is guided by a fiber optic **104** and a light guide **106** into the detector **18**. The photodiodes **26** are substantially insensitive to the UV light. Preferably, the photodiodes **26** are inherently UV blind. Alternatively, a UV blocking filter layer can be deposited on the photodiode surface. The ultraviolet light **102** is used as a reference illumination in place of the secondary x-ray source of FIGURE 4 for measuring the detector gain **A(t)**. More specifically, a UV light pipe **108** is mounted to an inner surface **110** of the scintillating layer **24**. The inner surface **110** is preferably painted with a reflective layer **112**. Small openings **114** in the reflective layer allow the UV light **102** to reach the scintillating layer **24** such that the light **102** spreads out to illuminate the scintillating crystals and reach the photodiodes **26** which produce electrical signals in proportion to the light. The signals are measured and corrected by the correction means **38** as it is explained above.

The UV light pipe **108** is substantially transparent to x-rays and could be made from any UV transparent material such as certain glasses, quartz, and the like. The light pipe **108** preferably is coated with a second layer **118** of a reflective material to keep the UV light **102** from escaping the pipe **108**. The UV intensity reaching each detector element need not be equal since a correction for unequal intensities can be made. In one embodiment, a single UV source **100** serves several detection modules using multiple fiber optics.

With reference to FIGURE 8, the x-ray to analog converting layer **22** includes an x-ray photoconductor **128** which is excited by the pulsed light source **100**. The light injected into the photoconductor **128** produces carriers which are similar to the carriers produced by the x-rays which are absorbed in the same material. The light source **100** produces the reference illumination used for the detector gain measurement. The x-ray photoconductor **128** includes a top electrode layer **130** and bottom or pixel electrodes **132**.

A high voltage power supply **134** supplies a high voltage bias to the detector **18** via the top electrode **130**. The light pipe **108** is disposed above the top electrode **130** to distribute the light **102** to the openings **136** in the top electrode layer **130** via the fiber optic **104** and the light guide **106**. The light pipe **108** is preferably substantially transparent to x-rays and could be made of optically clear glass, plastic or other like transparent materials. The bottom electrodes **132** are individual electrodes that detect the signals for several separate x-ray sensitive regions of the detector **18**. The signals are measured and corrected by the correction means **38** as it is explained above.

With reference to FIGURE 9, a charge carrier injecting means **138** injects carriers into the photoconductor **128**. Such artificially injected carriers behave in substantially the same manner as the carriers created by the x-rays. Trapping effects in the photoconductor, which affect the gain during x-ray detection, also affect the gain for carrier injection by electrodes. More specifically, for the purpose of measuring the detector gain **A(t)**, the top electrode layer **130** is partitioned into two or more electrodes, e.g. a common bias and injection top electrodes **140, 142**. Under regular scanning conditions, the top electrodes **140, 142** are at the same fixed potential. More specifically, when a switch **144** is in an “OFF” position, the high voltage power supply **134** supplies the high voltage bias to both top electrodes **140, 142**. During the measurement of the detector gain **A(t)**, the primary x-ray source **14** is turned OFF. The switch **144** is in an “ON” position. An injection power supply **146** supplies an injection potential to the injection electrodes **142** while the high voltage power supply **134** supplies the high voltage bias to the common bias electrodes **140**. In this manner, a differential voltage is applied to the common bias and injection electrodes **140, 142**. The differential voltage causes injection of carriers into the photoconductor **128** at the top surface **110**. The carriers create a signal which is detected by the bottom electrodes **132**. The signal is measured and corrected by the correction means **38** as it is explained above.

To prevent unwanted carriers from being injected into the photoconductor **128**, a thin layer **146**, also called a blocking contact, is disposed between the metal top electrodes **140, 142** and the x-ray photoconductor **128**. Injection of the unwanted carriers results in high dark current which is a highly undesirable effect. The blocking contact **146** can be formed by different methods such as P-N junction, a Schottky barrier, an insulating-tunneling barrier, and the like. Under normal conditions, when the high voltage bias is applied, the blocking contact **146** prevents injection of the carriers. The application of an extra high differential potential to the top electrodes **140, 142** causes a non-destructive breakdown of the blocking contact **146** with a resulting injection of carriers. Each of the three types of blocking contacts mentioned above is prone to this breakdown.

The use of a pulsed x-ray source in combination with the above described mathematical correction will allow detectors to be used for which gain and offset vary with time. The examples of such previously unusable detectors are detectors made of low grade scintillators, x-ray photoconductors, and the like.

In one embodiment, a statistical history of changes in the detector's offset and/or gain with time is collected. The system is calibrated less frequently based on the collected history.

The invention has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

CLAIMS

Having described the preferred embodiments, the invention is now claimed to be:

1. A radiographic imaging apparatus **(10)** comprising:

a primary radiation source **(14)** which projects a beam of radiation into an examination region **(16)** in which a subject **(28)** is disposed for an examination;

a detector **(18)** which converts detected radiation passing through the examination region **(16)** into electrical detector signals representative of the detected radiation and at least one temporally changing characteristic; and

a correction means **(38)** for determining a correction to the detector signals to compensate for the at least one temporally changing characteristic a plurality of times during generation of the detector signals, and correcting the detector signals with the determined corrections.

2. The apparatus as set forth in claim 1, wherein the changing characteristic is an offset **B(t)** and the correction means **(38)** includes an offset means **(84)** for measuring the offset **B(t)** of the detector **(18)** and further including:

a grid pulse means **(64)** for fast pulsing the primary radiation source **(14)**, the offset **B(t)** staying substantially constant during a primary radiation source ON time **(t_1)** and a primary radiation source OFF time **(t_2)**, which offset means **(84)** measures a value of the offset **(B_2)** at the primary source OFF time **(t_2)**, which offset value **(B_2)** is equal to a signal value **(S_2)** during the OFF time **(t_2)**.

3. The apparatus as set forth in claim 2, wherein the grid pulse means **(64)** turns the primary radiation source **(14)** ON and OFF at a rate between 1000 and 5000 pulses per second, such that the offset **B(t)** is re-measured between 1000 and 5000 times per second.

4. The apparatus as set forth in claim 2, wherein the correction means **(38)** corrects the detected signal by subtracting the measured offset value **(B_2)** from the detected signal:

$$S_{C1} = A_1X_1 - B_2\text{, where}$$

$\mathbf{S_{C1}}$ is the corrected signal during the time **(t_1)**,

$\mathbf{A_1}$ is a gain of the detector **(18)** during the time **(t_1)**,

$\mathbf{X_1}$ is an incident x-ray intensity on the detector **(18)** during the time **(t_1)**, and

$\mathbf{B_2}$ is the measured offset of the detector **(18)** during the time **(t_2)**.

5. The apparatus as set forth in claim 2, further including:

a second pulsed source **(86, 100, 138)** of a constant intensity **(X_{Ref})** for exciting the detector **(18)**; and

a second pulse means **(88)** for fast pulsing the second source **(86, 100, 138)**.

6. The apparatus as set forth in claim 5, wherein the correction means **(38)** further includes:

a calibration means **(92)** for determining a nominal gain **(A_{Cal})** of the detector **(18)** which is determined in relation to a strength of a signal **(S_A)** measured when the second source **(86, 100, 138)** is ON during a time **(t_A)** and a signal **(S_B)** measured when the second source **(86, 100, 138)** is OFF during a time **(t_B)**:

$$A_{Cal} = [S_A - S_B]/ X_{Ref}$$

$\mathbf{A_{Cal}}$ is the calibration gain of the detector **(18)**,

$\mathbf{S_A}$ is the measured signal of the detector **(18)** during the time **(t_A)**,

$\mathbf{X_{Ref}}$ is the intensity of the secondary source **(86)**, and

$\mathbf{S_B}$ is the measured signal of the detector **(18)** during the time **(t_B)**.

7. The apparatus as set forth in claim 6, wherein the correction means **(38)** includes a gain means **(94)** for correcting the detected signal in accordance with a change in a gain **(A_t)** of the detector **(18)** as:

$$S_{C1} = \{[S_A - S_B] \bullet [S_1 - S_2]\} / \{S_3 - S_2\}\text{, where}$$

$\mathbf{S_{C1}}$ is a corrected signal of the detector **(18)** during the time **(t_1)**,

S_A is the measured signal of the detector **(18)** during the time **(t_A)**,
S_B is the measured signal of the detector **(18)** during the time **(t_B)**,
S_1 is the measured signal of the detector **(18)** during the time **(t_1)**,
S_2 is the measured signal of the detector **(18)** during the time **(t_2)**,
S_3 is a measured signal of the detector **(18)** during a time **(t_3)**,
wherein the time **(t_3)** is an ON time of the second source **(86)** while the primary source **(14)** is OFF, which time **(t_3)** is close in time to the times **(t_1)** and **(t_2)**.

8. The apparatus as set forth in claim 7, wherein the converting layer **(22)** includes:

an array **(24)** of scintillating crystals which crystals detect radiation events and convert the detected events into light; and

an array of photodiodes **(26)** which photodiodes **(26)** are coupled to the scintillating crystals to detect light and convert the detected light into electrical signals.

9. The apparatus as set forth in claim 7, wherein the converting layer **(22)** includes:

an array **(128)** of x-ray photoconductors which photoconductors detect radiation events and convert the detected radiation events into electrical signals.

10. The apparatus as set forth in claim 5, wherein the second pulsed source is one of an x-ray source **(86)**, a light source **(100)** and a carrier injecting source **(138)**.

11. The apparatus as set forth in claim 5, wherein the second pulsed source is a UV light source.

12. The apparatus as set forth in claim 1, wherein the changing characteristic is at least one of an offset **B(t)** and a gain **A(t)** and further including:

a rotating gantry **(32)** which rotates the primary radiation source **(14)** around an axial direction **(Z)** to project the radiation beam at a variety of angular positions;

a second pulsed source **(86, 100, 138)** of a constant intensity **(X_{Ref})** for exciting the detector **(18)** which second source is disposed about the examination region;

a grid pulse means **(64)** for rapidly pulsing the primary source **(14)**; and

a second pulse means **(88)** for rapidly pulsing the second source **(86, 100, 138),** wherein the correction means **(38)** re-measures and corrects at least one of the offset **B(t)** and gain **A(t)** at a rate of pulsing of corresponding pulse means **(64, 88)**.

13. The apparatus as set forth in claim 12, wherein an intensity of the primary source **(14)** is adjusted by the grid pulse means **(64)** in accordance with an angular position of the primary x-ray source **(14)** to achieve a fast dose modulation.

14. The apparatus as set forth in claim 1, further including:

an A/D converter **(34)** which converts analog electrical signals into a series of digital numbers and wherein the correction means **(38)** is at least partially integrated with the A/D converter **(34)**.

15. A method of radiographic imaging comprising:

projecting a beam of radiation into an examination region **(16)** in which a subject **(28)** is disposed for an examination;

converting detected radiation passing through the examination region **(16)** into electrical detector signals representative of the detected radiation and at least one temporally changing characteristic;

determining a correction to the detector signals to compensate for the at least one temporally changing characteristic a plurality of times during generation of the detector signals; and

correcting the detector signals with the determined corrections.

16. The method as set forth in claim 15, wherein the changing characteristic is an offset **B(t)** and the correction step includes measuring the offset of the detector **(18)** and further including:

fast pulsing a primary radiation source **(14)**, the offset **B(t)** staying substantially constant during a primary radiation source ON time **(t_1)** and a primary radiation source OFF time **(t_2)**; and

measuring a value of the offset **(B_2)** at the primary source OFF time **(t_2)**, which offset value **(B_2)** is equal to a signal value **(S_2)** during the OFF time **(t_2)**.

17. The method as set forth in claim 16, wherein the step of pulsing includes:

turning the primary radiation source **(14)** ON and OFF at a rate between 1000 and 5000 pulses per second; and

measuring the offset between 1000 and 5000 times per second.

18. The method as set forth in claim 16, further including:

exciting the detector **(18)** with a second pulsed source **(86, 100, 138)** of a constant intensity; and

fast pulsing the second source **(86, 100, 138)**.

19. The method as set forth in claim 18, wherein the correction step further includes:

determining a nominal gain **(A_{Cal})** of the detector **(18)** in relation to a strength of a signal **(S_A)** measured when the second source is ON during a time **(t_A)** and the signal **(S_B)** measured when the second source is OFF during a time **(t_B)**:

$$A_{Cal} = [S(t_A) - S(t_B)] / X_{Ref}$$

A_{Cal} is a calibration gain of detector **(18)**,
S_A is a measured signal of the detector **(18)** during the time **(t_A)**,
X_{Ref} is an intensity of the secondary source **(86)**, and
S_B is a measured signal of the detector **(18)** during the time **(t_B)**.

20. The method as set forth in claim 19, wherein the correction step further includes:

correcting the detected signal in accordance with a change of a gain **(A_t)** of the detector as:

$$S_{C1} = \{[S_A - S_B] \bullet [S_1 - S_2]\} / \{S_3 - S_2\}, \text{ where}$$

S_{C1} is a corrected signal of the detector **(18)** during the time **(t_1)**,

S_A is the measured signal of the detector **(18)** during the time **(t_A)**,
S_B is the measured signal of the detector **(18)** during the time **(t_B)**,
S_1 is a measured signal of the detector **(18)** during the time **(t_1)**,
S_2 is a measured signal of the detector **(18)** during the time **(t_2)**,
S_3 is a measured signal of the detector during a time **(t_3)**,
wherein the time **(t_3)** is an ON time of the second source **(86)** while the primary source **(14)** is OFF, which time **(t_3)** is close in time to the times **(t_1)** and **(t_2)**.

21. The method as set forth in claim 20, wherein the detecting step includes:

detecting radiation events and converting the detected radiation events into electrical signals with an array **(128)** of x-ray photoconductors.

22. The method as set forth in claim 16, further including:

during at least some of the OFF times, exciting the detectors with one of lower energy x-ray radiation, light, and injected charge carriers.

23. The method as set forth in claim 15, wherein the changing characteristic is at least one of an offset **B(t)** and a gain **A(t)** and further including:

rotating the primary radiation source **(14)** around an axial direction **(Z)** to project the radiation beam at a variety of angular positions;

exciting the detector **(18)** with a second pulsed source **(86, 100, 138)** of a constant intensity **(X_{Ref})**;

rapidly pulsing the primary source **(14)**;

rapidly pulsing the second source **(86, 100, 138)**; and

re-measuring and correcting at least one of the offset **B(t)** and gain **A(t)** at a rate of pulsing of corresponding pulse means **(64, 88)**.

24. The method as set forth in claim 23, further including:

continually adjusting an intensity of the primary source **(14)** in accordance with an angular position of the primary source to achieve a fast dose modulation.

25. A computed tomography scanner for performing the method of claim 15.

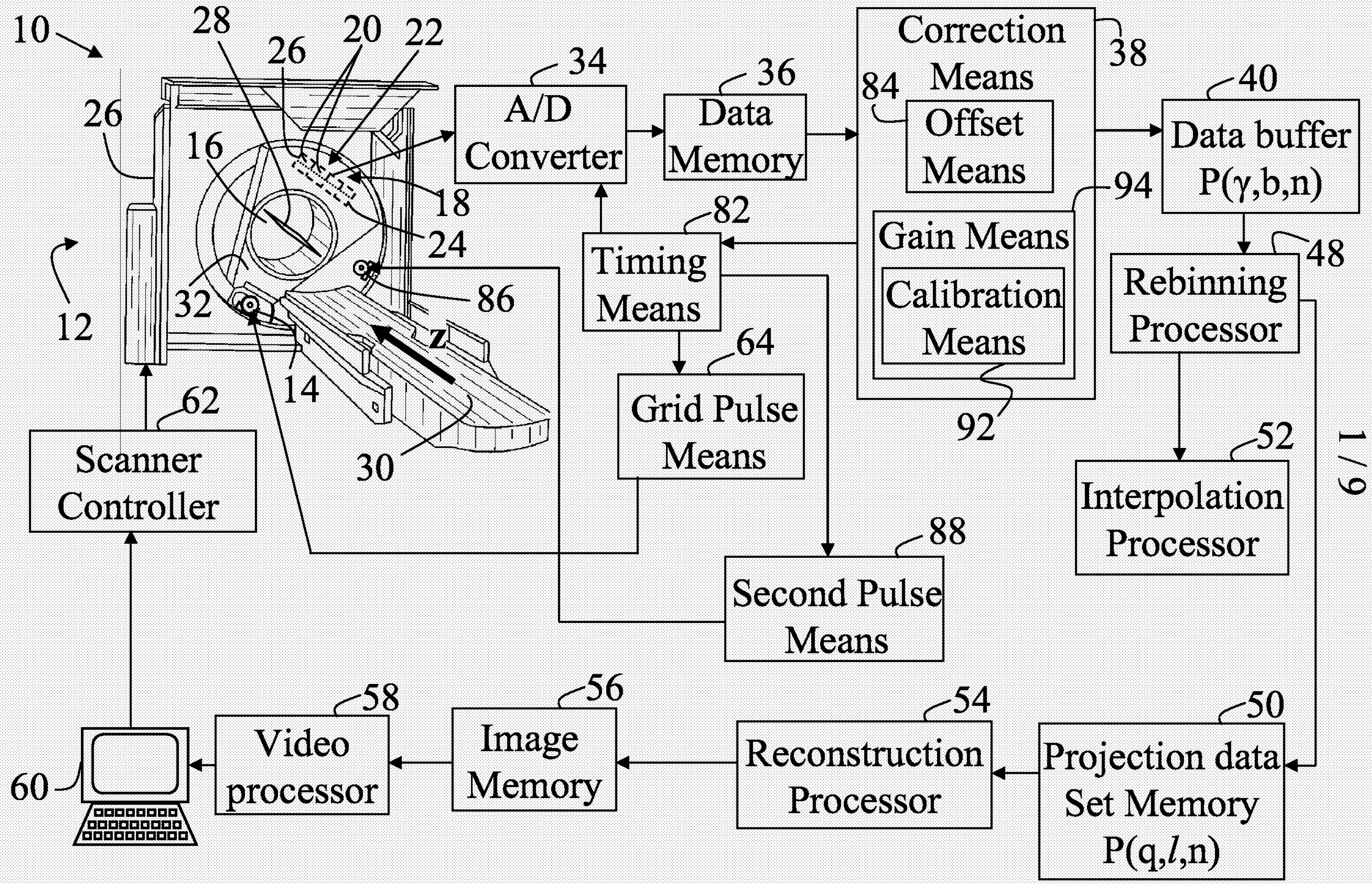
10
12
14
16
18
20
22
24
26
26
28
30
32
Z
86
34
A/D Converter
36
Data Memory
38
Correction Means
84
Offset Means
94
Gain Means
Calibration Means
92
40
Data buffer P(γ,b,n)
48
Rebinning Processor
52
Interpolation Processor
82
Timing Means
64
Grid Pulse Means
88
Second Pulse Means
62
Scanner Controller
60
58
Video processor
56
Image Memory
54
Reconstruction Processor
50
Projection data Set Memory P(q,l,n)

FIG 1

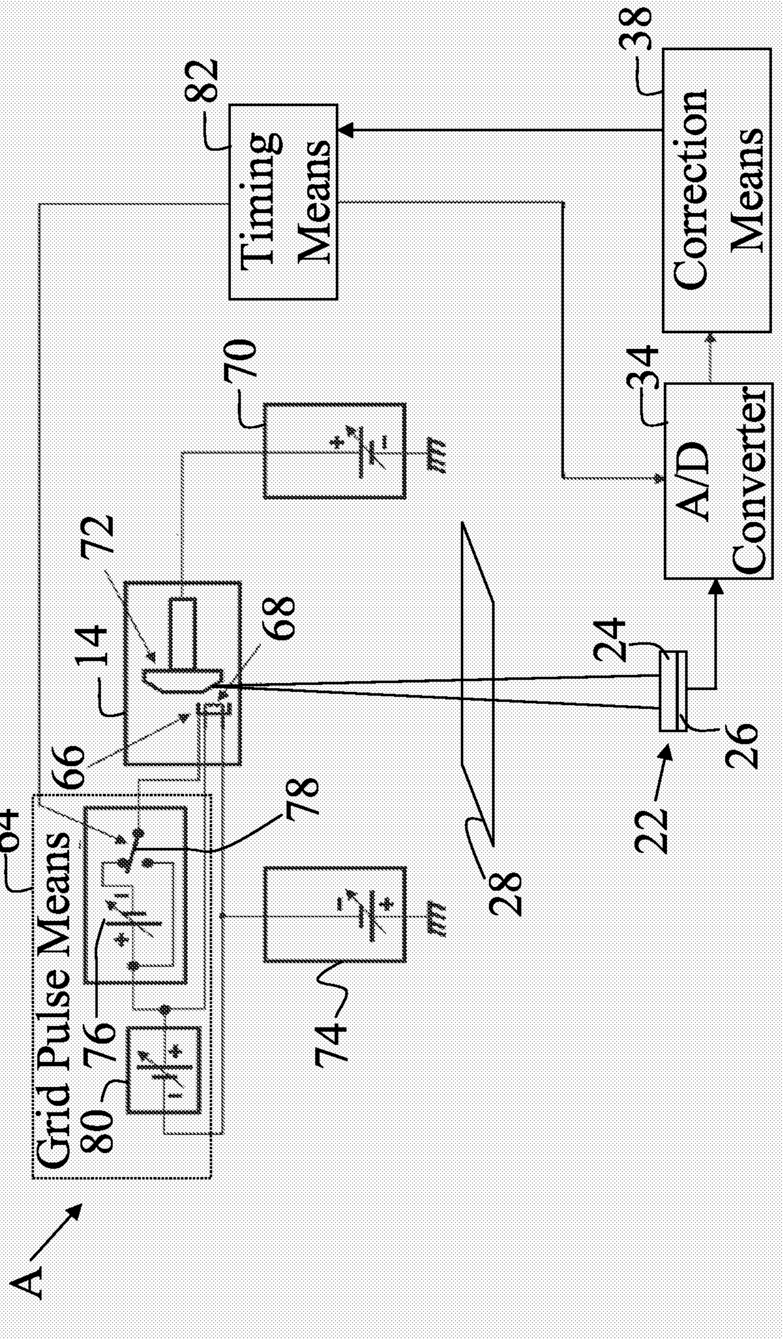
A
Grid Pulse Means
64
80
76
78
66
14
72
68
70
74
28
22
24
26
A/D
Converter
34
Correction
Means
38
Timing
Means
82

FIG 2

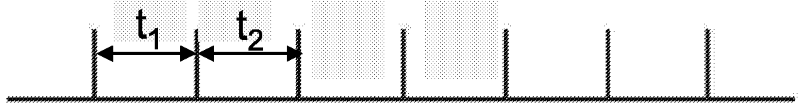

FIG 3A

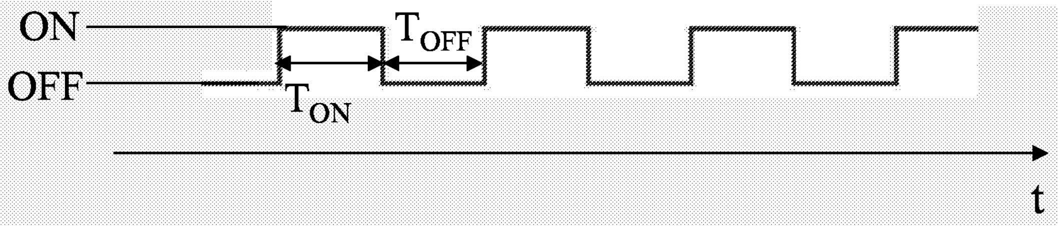

FIG 3B

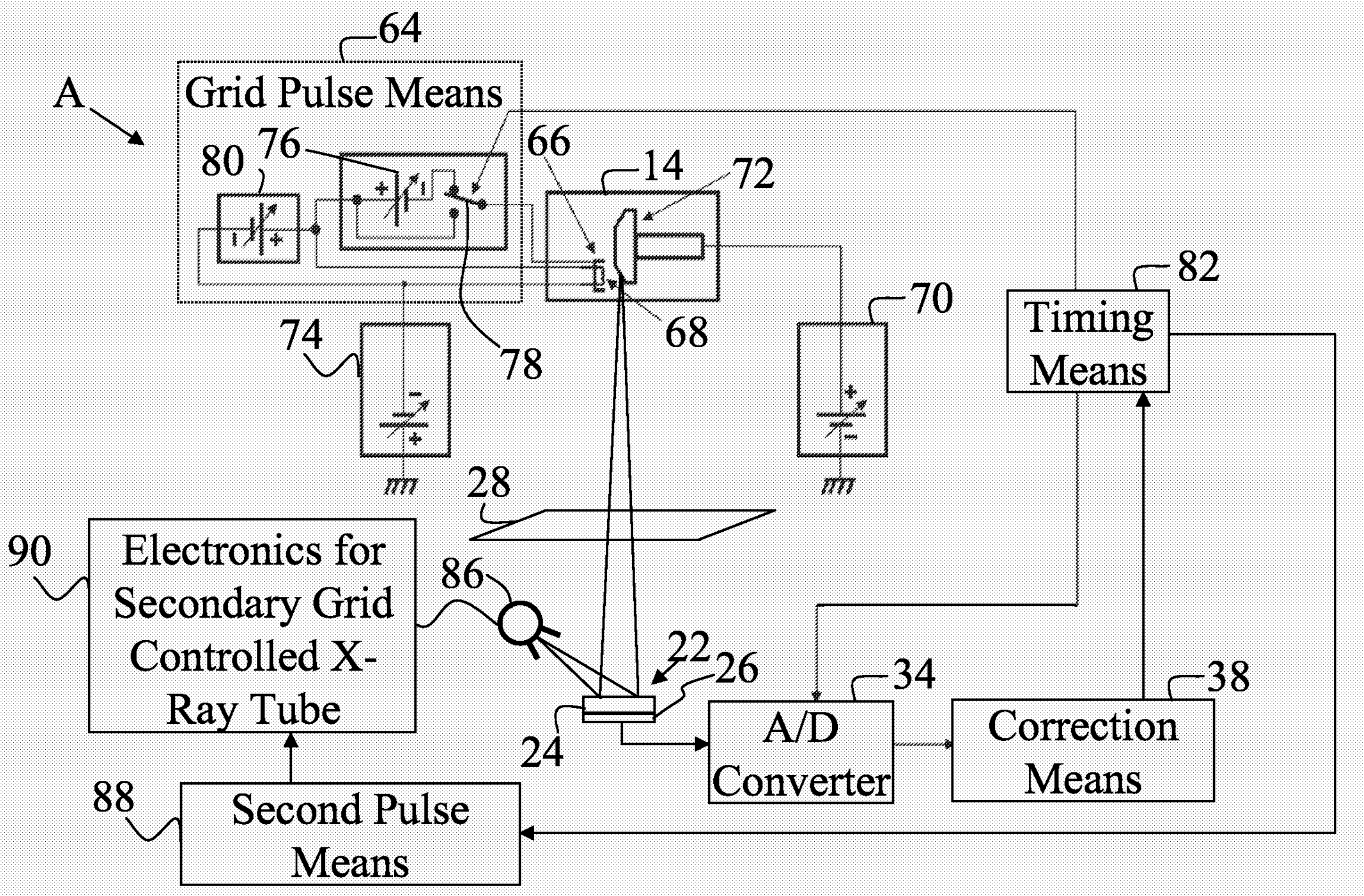
A
64
Grid Pulse Means
80
76
66
14
72
74
78
68
70
82
Timing Means
28
90
Electronics for Secondary Grid Controlled X-Ray Tube
86
22
26
34
38
24
A/D Converter
Correction Means
88
Second Pulse Means

FIG 4

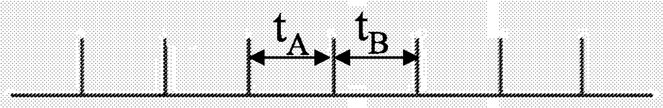

FIG 5A

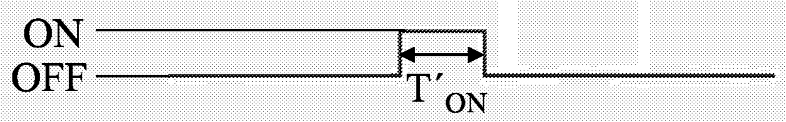

FIG 5B

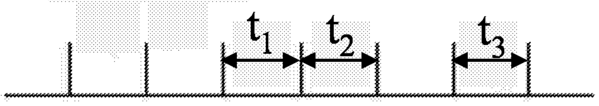

FIG 5C

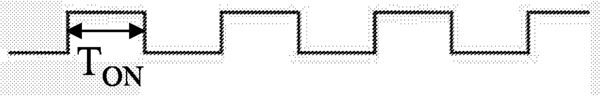

FIG 5D

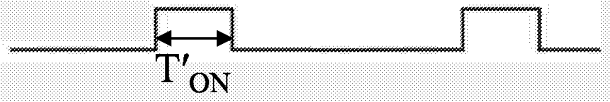

FIG 5E

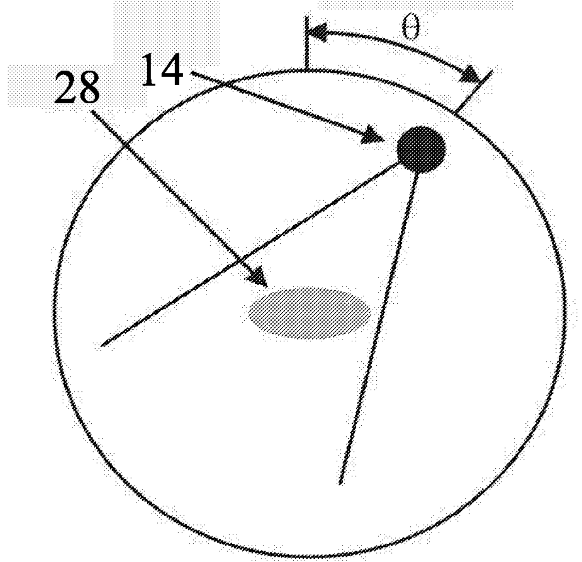

FIG 6A

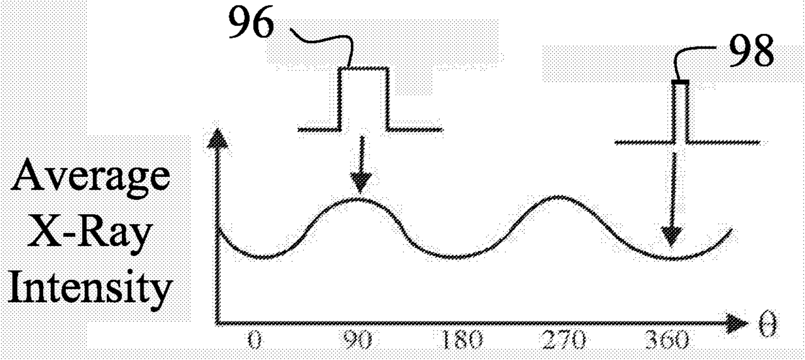

FIG 6B

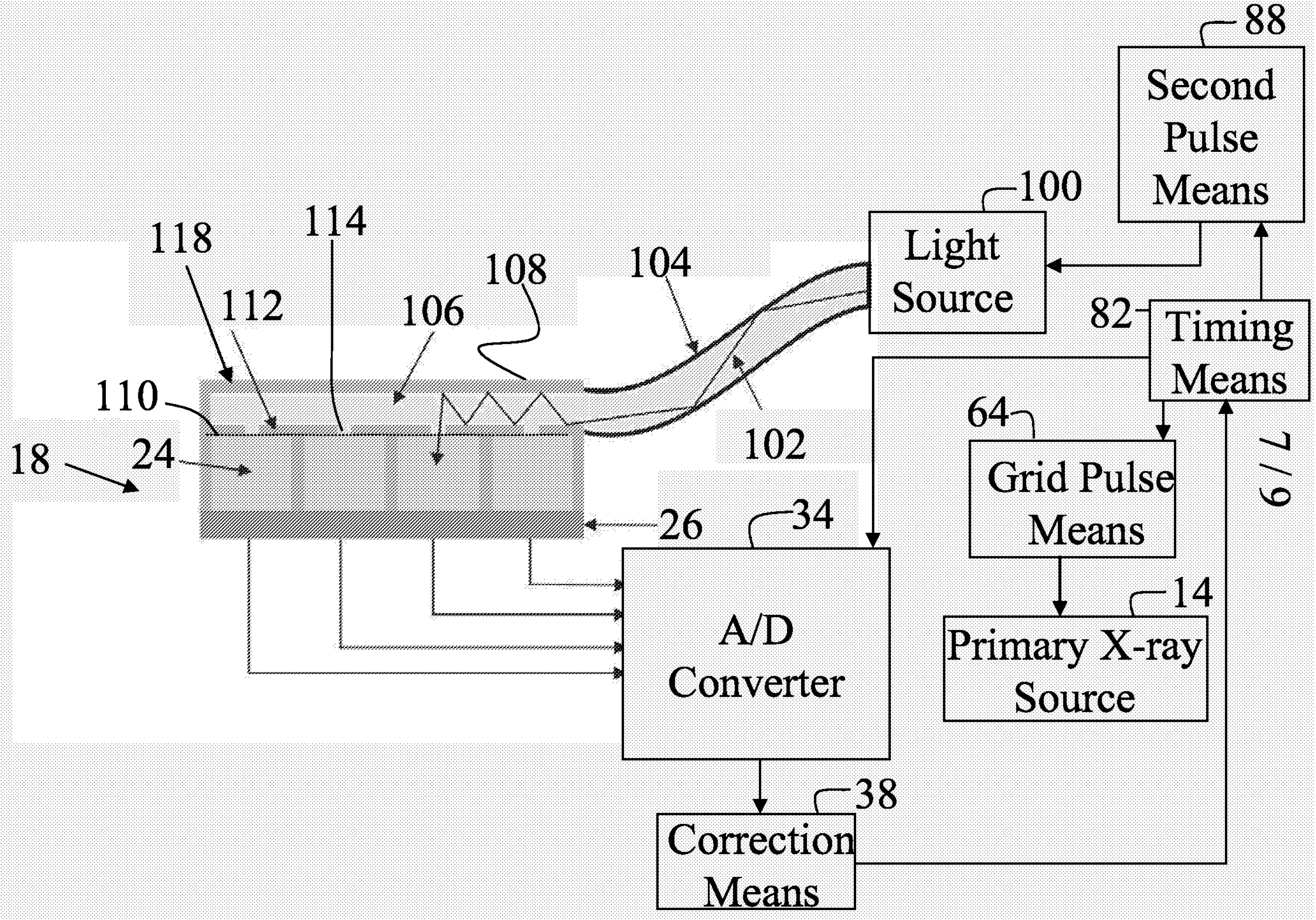
118
114
112
108
104
106
110
18
24
102
100
Light Source
88
Second Pulse Means
82
Timing Means
64
Grid Pulse Means
14
Primary X-ray Source
26
34
A/D Converter
38
Correction Means

FIG 7

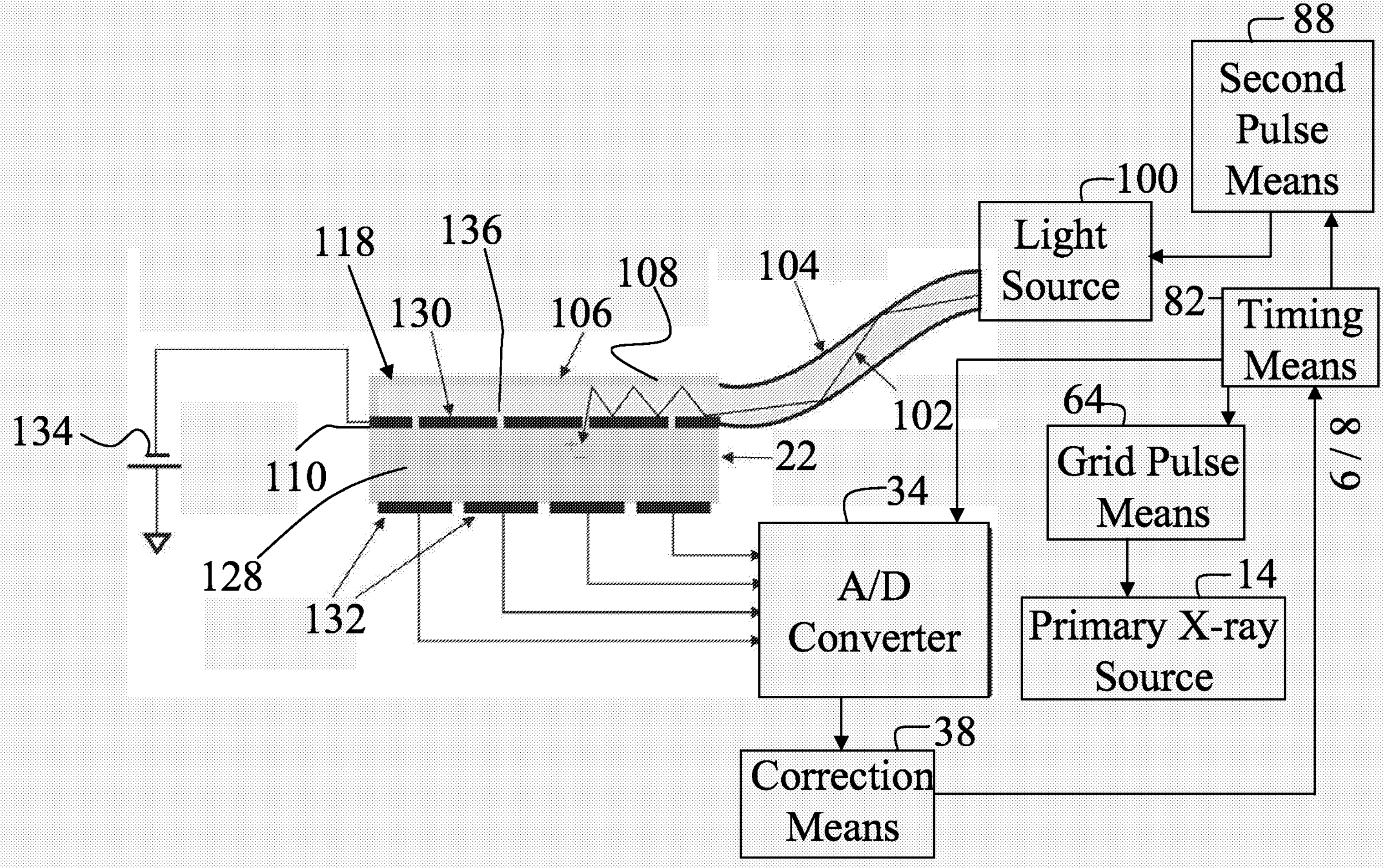
88
Second Pulse Means
100
Light Source
82
Timing Means
64
Grid Pulse Means
14
Primary X-ray Source
34
A/D Converter
38
Correction Means
118
136
130
106
108
104
102
134
110
22
128
132

FIG 8

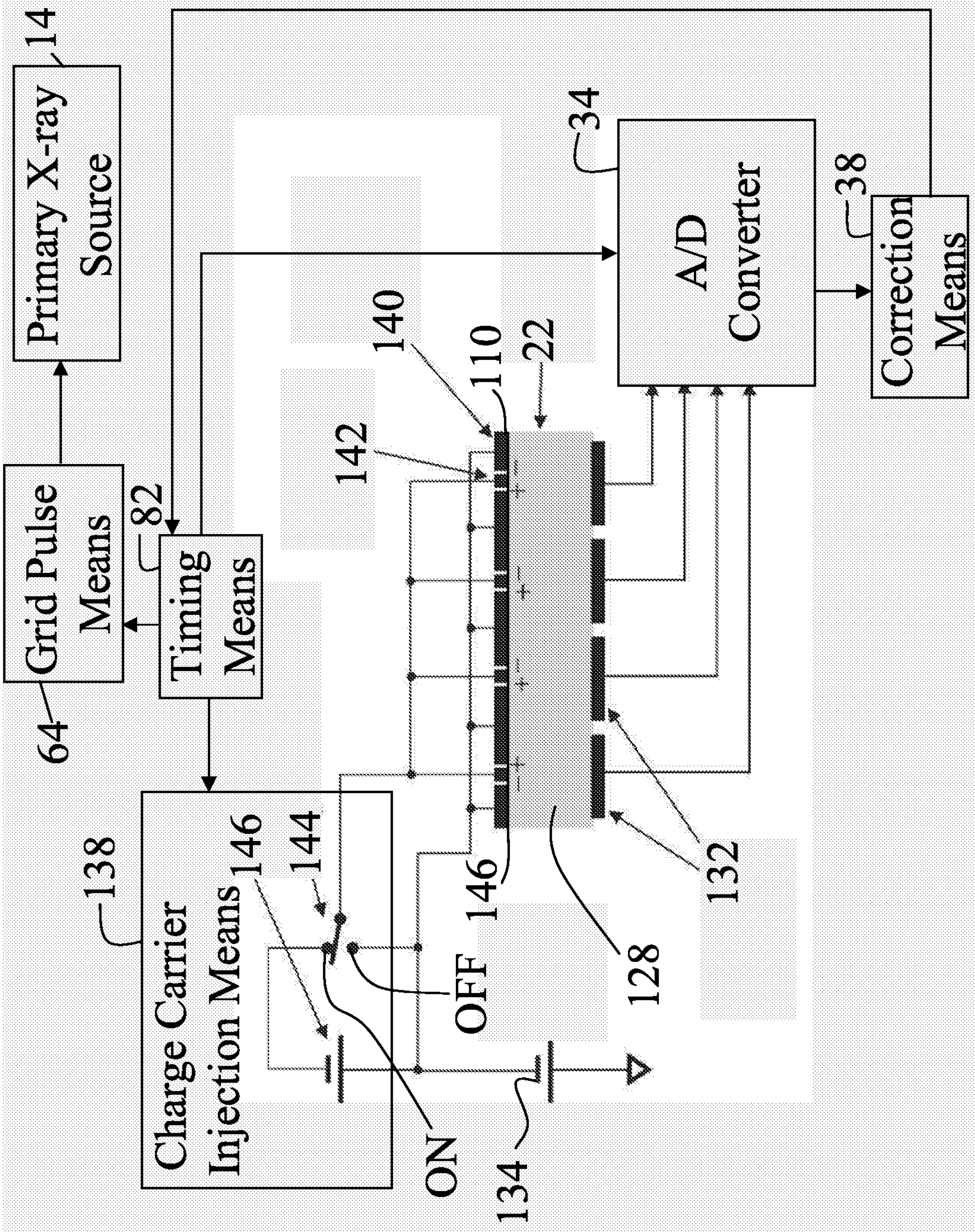
Primary X-ray Source
14
Grid Pulse Means
64
Timing Means
82
Charge Carrier Injection Means
138
146
144
OFF
ON
134
140
142
110
22
146
128
132
A/D Converter
34
Correction Means
38

FIG 9