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X-RAY SENSOR SIGNAL PROCESSOR AND X-RAY COMPUTED
TOMOGRAPHY SYSTEM USING THE SAME

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BACKGROUND OF THE INVENTION

The present invention relates to an X-ray sensor signal processor for processing output signals of X-ray sensors for non-destructive inspection, and an X-ray computed tomography system (hereinafter referred to as "X-ray CT system") using the X-ray sensor signal processor. Particularly, it relates to an X-ray sensor signal processor in which a dark current generated in each of X-ray sensors can be removed and a value proportional to the average number of photons in X-rays can be obtained even in the case where a small number of incident photons are given, and an X-ray CT (computed tomography) system using the X-ray sensor signal processor.

Industrial X-ray CT systems using X-ray pulses with high energy (of 1 MeV or higher) have been developed in recent years for the purpose of non-destructively inspecting internal defects, or the like, of metallic parts or devices. Greater progress in research and development has been made for the purpose of inspecting a larger object with high resolution.

An X-ray CT system has been described in H. Miyai, et al. "A High Energy X-Ray Computed Tomography Using Silicon Semiconductor Detectors", 1996 Nuclear Science Symposium Conference Record, Vol. 2, pp 816-820,

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Nov. 2-9, 1996, Anaheim, CA, USA (1997) (hereinafter referred to as "first background art"). A signal processor for processing output signals of X-ray sensors shown in the first background art will be described

5 below with reference to Fig. 9.

In the signal processor 1 shown in Fig. 9, semiconductor sensors (X-ray sensors) 21 to 2n are connected to first-stage circuits 90 to 9n respectively. Because a semiconductor sensor for an X-ray computed tomography is shaped like a strip of paper with a large size (for example, $3 \times 40 \times 0.4$ mm) to detect a high-energy X-ray pulse efficiently, there is a possibility that a dark current with a high level of the order of tens of nA may be inevitably generated in the semiconductor sensor. In the first-stage circuit 90, a capacitor 114 is provided to AC-couple the semiconductor sensor 21 so that a voltage amplifier does not amplify a DC voltage caused by the dark current. When X-rays become incident onto the semiconductor sensor 21, an electric current flows through a capacitor 114 because electric charges are generated in the inside of the sensor. A voltage change generated on this occasion is amplified by two stages of voltage amplifiers 92 and 92', held by a sample/hold amplifier 94 and supplied to a posterior device.

Fig. 10A shows an output of the semiconductor sensor 21 in the first background art, and Fig. 10B shows an output of the sample/hold amplifier 94 in the

first background art. An X-ray pulse is output with a pulse width T_w . Hence, when there is no object or when an object penetrated by the X-ray pulse is thin, the output of the X-ray sensor is provided as a rectangular-wave output 101 (solid line) proportional to the dose of the X-ray pulse as shown in Fig. 10A. Hence, the output of the sample/hold amplifier 94 is obtained as an output 103 proportional to the average number of photons per unit time except the first rising portion as shown in Fig. 10B.

On the contrary, when an X-ray pulses pass through a thick object, the number of incident photons is reduced by at least four digits compared with the case where no object is set, because the X-ray pulses are attenuated before the X-ray pulses become incident on the semiconductor sensor 21. The output 102 in Fig. 10A shows an example of the output of the semiconductor sensor 21 in the case where the object is so thick that a small number of incident photons are given. As shown in Fig. 10A, the output of the semiconductor sensor 21 is not kept constant but it has a waveform the output height of which is heightened only when photons are incident on the semiconductor sensor 21. In this case, the output of the sample-and-hold amplifier 94 is not kept constant, either, as represented by the output 104 in Fig. 10B. That is, the value held by the sample/hold amplifier 94 does not always express a voltage proportional to the average number of photons per unit time.

As described above, the first background art has a problem that a voltage proportional to the average number of photons per unit time is not obtained when a small number of incident photons are given.

5 A technique for obtaining a voltage proportional to the average number of photons even in the case where a small number of incident photons are given has been described in JP-A-58-15847 (hereinafter referred to as "second background art"). In the second background
10 art, there has been described a medical X-ray tomography system in which a voltage proportional to the average number of photons is obtained by a root mean square of fluctuation components obtained as a result of removal of a DC component from an output signal of an X-ray
15 sensor by a capacitor. This technique is called "root mean square voltage technique" or "Campbell's technique".

 Assume now the case where the second background art is applied to the first background art.
20 Generally, the pulse width of an X-ray pulse used in the industrial X-ray CT system which is a subject of the first background art is set to be small (about 5 μ s) compared with the medical X-ray tomography system which is a subject of the second background art. The pulse
25 with such a small pulse width, however, cannot be removed by a capacitor. Hence, fluctuation components cannot be taken out exclusively by a capacitor even in the case where the second background art is applied to

the industrial X-ray CT system which is a subject of the first background art. Because a voltage proportional to the average number of photons cannot be obtained accurately by the root mean square voltage technique if
5 fluctuation components cannot be taken out exclusively, a voltage proportional to the average number of photons cannot be obtained accurately even in the case where the second background art is applied to the first background art as described above.

10 SUMMARY OF THE INVENTION

An object of the present invention is to provide an X-ray sensor signal processor in which a dark current generated in each X-ray sensor can be removed and a value proportional to the average number of
15 photons in X-rays can be obtained even in the case where a small number of incident photons are given, and an X-ray CT system using the X-ray sensor signal processor.

The present invention is characterized by an X-ray sensor signal processor for processing an output
20 signal of each X-ray sensor for detecting pulse-like X-rays emitted from an accelerator and passed through an object, wherein the processor comprises a filter for removing a DC component from the output signal of the X-ray sensor, and an integrator for integrating the output
• 25 signal of the X-ray sensor after removal of the DC component by the filter.

According to the present invention, a dark

current generated in the X-ray sensor can be removed because the capacitor for removing a DC component from the output signal of the X-ray sensor is provided, and a value proportional to the average number of photons in
5 X-rays can be obtained by integration of the output signal of the X-ray sensor even in the case where a small number of incident photons are given because the integrator for integrating the output signal of the X-ray sensor after removal of the DC component is
10 provided.

According to the present invention, a dark current generated in each of X-ray sensors can be removed and a value proportional to the average number of photons can be obtained even in the case where a
15 small number of incident photons are given.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 shows a configuration diagram of a signal processor as a preferred embodiment of the present invention.

20 Figs. 2A and 2B show configuration diagrams of an X-ray CT system as a preferred embodiment of the present invention.

Figs. 3A and 3B show graphs showing the output current waveform of the semiconductor sensor 21 depicted
25 in Fig. 1 and the output voltage waveform of the integrator depicted in Fig. 1.

Fig. 4 shows a flow chart showing the

operation of the control circuit 12 depicted in Fig. 1.

Fig. 5 shows a configuration diagram of a signal processor as another embodiment of the present invention.

5 Fig. 6 shows a configuration diagram of a signal processor as a further embodiment of the present invention.

Fig. 7 shows a configuration diagram of a signal processor as a further embodiment of the present
10 invention.

Fig. 8 shows a configuration diagram in the case where five signal processors 1a to 1e are used in the embodiment of Fig. 1.

Fig. 9 shows a configuration diagram of a
15 signal processor in the first background art.

Figs. 10A and 10B show graphs showing the output current waveform of the semiconductor sensor 21 depicted in Fig. 9 and the output voltage waveform of the sample/hold amplifier 94 depicted in Fig. 9.

20 DETAILED DESCRIPTION OF THE INVENTION

Embodiments of the present invention will be described below in detail by using the drawings.

(Embodiment 1)

Fig. 1 shows an X-ray sensor signal processor
25 which is a preferred embodiment of the present invention. Fig. 2 shows an X-ray CT system using the X-ray sensor signal processor depicted in Fig. 1.

Referring first to Fig. 2, non-destructive inspection by the X-ray CT system will be described. Fig. 2A is a side view of the X-ray CT system, and Fig. 2B is a partial top view of the X-ray CT system. In
5 Fig. 2A, first, a CT controller 9 issues an X-ray pulse output command to an accelerator 4. At the same time when the CT controller 9 issues the X-ray pulse output command, the CT controller 9 issues an X-ray pulse output start signal to a signal processor (X-ray sensor
10 signal processor) 1. Upon reception of the X-ray pulse output command, the accelerator 4 generates fan beam-like (radial) X-rays with high energy (not lower than 1 MeV) by irradiation of a target 42 with an ion beam acceleratively emitted from a linear accelerator 41
15 (hereinafter referred to as "LINAC"). The generated X-rays are emitted from the accelerator 4 through a collimator 43. Although the X-rays are emitted as a pulse with a short pulse width of 5 μ s from the accelerator 4, both energy and pulse width of the ion
20 beam irradiating the target 42 are controlled by the LINAC 41 so that both energy and pulse width of the X-rays are controlled.

The X-ray pulse emitted from the accelerator 4 irradiates an inspection object 5 disposed on a scanner
25 6 and passes through the object 5. A control signal not shown is given to the scanner 6 by the CT controller 9, so that the scanner 6 rotates and moves up and down in accordance with the control signal. By the rotation and

up-down movement of the scanner 6, any portion of the object 5 is irradiated with X-ray pulses from various directions by a plurality of times. The X-ray pulses passed through the object 5 are incident on an array of semiconductor sensors (X-ray sensors) 21 to 2n via the collimator 7. When the X-rays are incident on a depletion layer of a p-n junction in any one of the semiconductor sensors 21 to 2n, a large number of electron-hole pairs are generated so that an electric current flows in the semiconductor sensor. The signal processor 1 issues sensor output data corresponding to the electric currents generated in the semiconductor sensors 21 to 2n. The operation of the signal processor 1 will be described later.

15 A data transmitter-receiver circuit 3 receives the sensor output data from the signal processor 1 and transmits the sensor output data to the CT controller 9. Incidentally, data transfer between the data transmitter-receiver circuit 3 and the CT controller 9 is performed through a cable passing through a shield wall 8 because there is the shield wall 8 between the data transmitter-receiver circuit 3 and the CT controller 9. Both data transfer between the accelerator 4 and the CT controller 9 and data transfer between the scanner 6 and the CT controller 9 are performed through the cable in the same manner as described above. The CT controller 9 reconstructs a perspective image of a section of the object 5 by using

the sensor output data given to the CT controller 9 and makes a display unit 10 display the perspective image.

In the aforementioned manner, a perspective image of a section of the object 5 is obtained.

5 Referring next to Fig. 1, the signal processor 1 will be described. In Fig. 1, the semiconductor sensors 21 to 2n are connected to first-stage circuits 111 to 11n respectively. Incidentally, only one first-stage circuit 111 will be described hereafter because
10 each of the first-stage circuits 112 (not shown) to 11n has the same configuration as that of the first-stage circuit 111.

In the first-stage circuit 111, a bias supply 118 is connected to one end of the semiconductor sensor
15 21 in a direction of applying a reverse bias to the semiconductor sensor 21, and a resistor 119 is connected to the other end of the semiconductor sensor 21. Incidentally, the other end of the resistor 119 is connected to the ground. Further, a capacitor 114 is
20 connected to a junction between the semiconductor sensor 21 and the resistor 119. The other end of the capacitor 114 is connected to an inverting input of an operational amplifier 115. A resistor 116 and a capacitor 117 are connected in parallel to the operational amplifier 115.
25 The operational amplifier 115, the resistor 116 and the capacitor 117 form an integrator. Hereinafter, the operational amplifier 115, the resistor 116 and the capacitor 117 will be collectively called "integrator".

The first-stage circuit 111 is configured as described above.

As described above in the background art, a dark current of the order of tens of nA is generated in the semiconductor sensor 21. This dark current is, however, a DC component. Hence, the dark current does not flow into the integrator through the capacitor 114 but flows into the resistor 119. In the first-stage circuit 111, the dark current is removed in this manner.

When an X-ray pulse is detected by the semiconductor sensor 21, the frequency band of the output current of the semiconductor sensor 21 is not lower than the order of tens of kHz because the pulse width of the X-ray pulse is 5 μ s. Hence, the output current flows into the integrator through the capacitor 114. Because the inverting input of the operational amplifier 115 forms an imaginary short circuit, the output current little flows into the resistor 116 but flows into the capacitor 117. Hence, electric charges are stored in the capacitor 117. When the X-ray pulse detection in the semiconductor sensor 21 is completed, the electric charges stored in the capacitor 117 are discharged (attenuated) in accordance with the time constant determined on the basis of the combination of the resistor 116 and the capacitor 117 because the output current from the semiconductor sensor 21 is merely constituted by the dark current as a DC component. Incidentally, a method for determining the

resistance R of the resistor 116 and the capacitance C of the capacitor 117, that is, a method for determining the time constant will be described later.

Fig. 3A shows the output current waveform of the semiconductor sensor 21, and Fig. 3B shows the output voltage waveform of the integrator constituted by the combination of the operational amplifier 115, the resistor 116 and the capacitor 117. In Fig. 3A, an output waveform 31 (solid line) shows the output waveform of the semiconductor sensor 21 in the case where an X-ray pulse not passed through the object 5 is incident on the semiconductor sensor 21, and the other output waveform 32 shows the output waveform of the semiconductor sensor 21 in the case where the number of incident photons in the X-ray pulse incident on the semiconductor sensor 21 is reduced to a one-digit number because the X-ray pulse passes through a thick portion of the object 5. As described above in the background art, the output waveform of the semiconductor sensor 21 is shaped to be a rectangular in proportion to the dose of the X-ray pulse in the case where the X-ray pulse does not pass through the object 5 whereas the output waveform of the semiconductor sensor 21 is shaped so as to be heightened only when photons are incident on the semiconductor sensor 21 in the case where the X-ray pulse passes through a thick portion of the object 5.

As shown in Fig. 3B, in the case where the X-ray pulse does not pass through the object 5, the

integrator output corresponding to the sensor output increases substantially linearly during the current output from the semiconductor sensor 21 and then electric discharge occurs gradually (output 33: solid
5 line). On the contrary, in the case where the X-ray pulse passes through a thick portion of the object 5, the output of the semiconductor sensor 21 is integrated so that a voltage proportional to the number of incident photons is obtained. That is, in the case where the X-
10 ray pulse passes through a thick portion of the object 5, a voltage proportional to the number of incident photons can be obtained in this embodiment because an integrated value is calculated by use of the integrator even if the output waveform of the semiconductor sensor
15 21 changes with the passage of time.

In this manner, according to the first-stage circuit 111 in this embodiment, a voltage proportional to the average number of photons in X-rays can be obtained accurately, whether the X-ray pulse passes
20 through the object 5 or not. Furthermore, no sample/hold amplifier needs to be provided in this embodiment because the integrator in this embodiment serves also as an sample/hold amplifier shown in the first background art. In addition, in this embodiment,
25 the bias state of the semiconductor sensor 21 or the capacitor 114 returns to a steady state immediately after the completion of the X-ray pulse irradiation because all electric charges generated by the X-ray

pulse are stored in the capacitor 117. Hence, no reset switch needs to be provided. In this manner, according to the first-stage circuit 111 in this embodiment, hardware configuration can be simplified. Incidentally,
5 as described above, each of the first-stage circuits 112 to 11n has the same configuration as that of the first-stage circuit 111.

Output ends of the first-stage circuits 111 to 11n (output ends of the integrators) are connected to
10 switches 131 to 13n, respectively, in a multiplexer 13. A control circuit 12 issues a "close" command to the switch 131 after 10 μ s from the start of X-ray pulse irradiation. Incidentally, an X-ray pulse emission start signal output from the CT controller 9 is supplied
15 to the control circuit 12 through the data transmitter-receiver circuit 3 and a bus 17. Hence, if irradiation starts when the X-ray pulse emission start signal is supplied to the control circuit 12, the control circuit 12 issues the "close" command after 10 μ s from the start
20 of irradiation. Upon reception of the "close" command, the switch 131 is closed so that the output signal of the first-stage circuit 111 is supplied to the A-D converter 14. The control circuit 12 issues "open" and "close" commands so that the switches 131 to 13n are
25 closed successively at intervals of 5 μ s.

Then, the control circuit 12 issues an A-D conversion command to the A-D converter 14. Upon reception of the A-D conversion command, the A-D

converter 14 converts the output signal of the first-stage circuit 111 into a digital output signal. The output signal obtained by the A-D conversion is stored in a memory 15 through the bus 17 by the control circuit 12. Incidentally, the memory 15 has storage regions which are provided in advance correspondingly to the semiconductor sensors 21 to 2n (first-stage circuits 111 to 11n) respectively. Hence, the aforementioned output signal is stored in a storage region corresponding to the semiconductor sensor 21.

Then, the control circuit 12 issues an "open" command to the switch 131 after 15 μ s from the start of X-ray pulse irradiation, that is, after 5 μ s from the issue of the "close" command to the switch 131. At the same time, the control circuit 12 issues a "close" command to the switch 132. Upon reception of the "open" command, the switch 131 is opened. Upon reception of the "close" command, the switch 132 is closed. As a result of the closing of the switch 132, the output signal of the first-stage circuit 112 is supplied to the A-D converter 14. Then, the control circuit 12 issues an A-D conversion command to the A-D converter 14, so that the A-D converter 14 converts the output signal of the first-stage circuit 112 into a digital output signal. The output signal obtained by the A-D conversion is stored in a storage region of the memory 15 corresponding to the semiconductor sensor 22 through the bus 17 by the control circuit 12.

The same processing as applied to the first-stage circuits 111 and 112 is hereafter applied to the other first-stage circuits 113 to 11n.

On this occasion, the output signal in each of
5 the first-stage circuits 111 to 11n is an output signal at a point of time when a predetermined time is passed after the completion of X-ray pulse irradiation. Hence, the level of the output signal at this point of time is smaller than that just after the completion of X-ray
10 pulse irradiation. That is, the level of the output voltage of the integrator in each of the first-stage circuits 111 to 11n is reduced with the passage of time because the output voltage is attenuated on the basis of the time constant determined by the product of the
15 resistance R of the resistor 116 and the capacitance C of the capacitor 117 as described above. In this embodiment, therefore, the output signal stored in the memory 15 is corrected so that an output signal just after the completion of X-ray pulse irradiation is
20 obtained.

The output signal level V_0 just after the completion of X-ray pulse irradiation is expressed by the following expression 1.

$$V_0 = V(t) \times \exp(t/\tau) \quad (\text{Expression 1})$$

In the expression 1, t is the time from the
25 completion of X-ray pulse irradiation, $V(t)$ is the

output signal level at time t , and τ is the time constant. In the case of the first-stage circuit 111, t is $5 \mu s$ because the time required for measurement of the output signal just after the completion of X-ray pulse irradiation is $5 \mu s$. Incidentally, t for each of the first-stage circuits 111 to 11n can be obtained in advance so that $t = 10 \mu s$ for the first-stage circuit 112, $t = 15 \mu s$ for the first-stage circuit 113, and so on, because the switches 131 to 13n are closed successively at intervals of $5 \mu s$ as described above. Also the time constant τ can be obtained in advance on the basis of the product of the resistance R of the resistor 116 and the capacitance C of the capacitor 117. Hence, the output signal level V_0 at the point of time just after the completion of X-ray pulse irradiation can be obtained on the basis of the expression 1.

In this embodiment, values are substituted for t and τ in $\exp(t/\tau)$ and $\exp(t/\tau)$ is stored, as a correction coefficient for corresponding one of the semiconductor sensors 21 to 2n (first-stage circuits 111 to 11n), in a correction coefficient memory 16 in advance. Each of the output signal levels stored in storage regions of the memory 15 corresponding to the semiconductor sensors 21 to 2n is corrected to the output signal level V_0 at the point of time just after the completion of X-ray pulse irradiation by the control circuit 12 with use of a corresponding correction coefficient stored in the correction coefficient memory

16 in advance (that is, the output signal level is multiplied by a corresponding correction coefficient). The corrected output signal levels V_0 are overwritten in storage regions of the memory 15 corresponding to the semiconductor sensors 2_1 to 2_n respectively. The output signal levels stored in the memory 15 are supplied, as sensor output data, to the data transmitter-receiver circuit 3 in response to the request of the CT controller 9.

10 The aforementioned processing is repeated whenever X-ray pulse irradiation is performed once. On this occasion, the output signals of the first-stage circuits 11_1 to 11_n , that is, the output signals of the integrators must be held until the output signal of the first-stage circuit 11_n is converted into a digital signal by the A-D converter 14 but the levels of the output signals must become zero until the next X-ray pulse irradiation. It is, therefore, necessary that the time constant for each integrator determined by the combination of the resistor 116 and the capacitor 117 is set to an appropriate value.

 A method for determining the resistance R of the resistor 116 and the capacitance C of the capacitor 117 in this embodiment will be described below. The following five matters must be considered when the resistance R of the resistor 116 and the capacitance C of the capacitor 117 are to be determined.

(1) $Q_{\max}$ [C]: maximum quantity of electric charges

generated by each of the semiconductor sensors 21 to 2n;

(2) m [digits]: dynamic range necessary for measurement;

(3) V_{fs} [V]: input range (maximum input voltage) of the A-D converter 14;

(4) n [bits]: required resolution; and

(5) T_p [s]: cycle period of X-ray pulse.

In the case where the X-ray pulse does not pass through the object 5, the output of each of the semiconductor sensors 21 to 2n reaches Q_{max} but the output voltage of corresponding one of the first-stage circuits 111 to 11n must not exceed the input range V_{fs} of the A-D converter 14. Further, the output signal level of the corresponding one of the first-stage circuits 111 to 11n must be attenuated to a value not higher than a voltage corresponding to the required resolution n in the time T_p (set to 5 ms in this embodiment) which is required before incidence of the next X-ray pulse.

The maximum output V_{max} of each of the integrators is determined by the following expression 2.

$$V_{max} = Q_{max}/C \quad (\text{Expression 2})$$

The relation $V_{max} \leq V_{fs}$ must be established because V_{max} over the input range V_{fs} of the A-D converter 14 cannot be measured.

Hence, the following expression 3 is deduced

from the expression 2.

$$C \geq Q_{\max}/V_{fs} \quad (\text{Expression 3})$$

Further, because the A-D converter 14 requires sufficient resolution to measure a smaller value by m digits than $V_{\max}$, the following expression 4 must be established if the attenuation of the signal before completion of the A-D conversion by the A-D converter 14 is neglected for the sake of simplification.

$$V_{\max} \times 10^{-m} = Q_{\max}/C \times 10^{-m} > V_{fs}/2^n \quad (\text{Expression 4})$$

On the other hand, the output of the integrator is attenuated on the basis of the time constant $\tau = R \times C$. Hence, the maximum output $V_{\max}$ of the integrator is attenuated as expressed by the following expression 5 at time t .

$$V(t) = V_{\max} \times \exp(-t/\tau) \quad (\text{Expression 5})$$

When measurement resolution of n [bits] is required, $V(t)$ must be attenuated to be not higher than a voltage corresponding to the required resolution in the time T_p before incidence of the next X-ray pulse. Hence, the following expression 6 must be satisfied.

$$V(T_p) = V_{\max} \times \exp(-T_p/\tau) \leq V_{\max}/2^n \quad (\text{Expression 6})$$

Accordingly, the time constant τ is given by the following expression 7.

$$\tau = R \times C \leq T_p / (n + 1 \ln 2) \quad (\text{Expression 7})$$

Further, from the relations among the expressions 3, 4 and 7, C is given by the following
5 expression 8.

$$Q_{\max} / V_{fs} \leq C \leq \min\{ (Q_{\max} \times 10^{-m} \times 2^n) / V_{fs}, T_p / (R(n + 1 \ln 2)) \}$$

(Expression 8)

Incidentally, the function "min()" means selection of smaller one of factors. In this embodiment, the cycle time T_p of X-ray pulses can be kept constant by trigger control of the accelerator because
10 this embodiment is applied to an industrial X-ray CT system. When accuracy of 14-bit resolution ($n = 14$) is required in use of a general A-D converter of 10 [V] full range ($V_{fs} = 10$ [V]), it is apparent from the expression 8 that C needs to be set in a range of 15
15 [pF] < C < 34 [pF] if the resistance R is 10 [M Ω] because the maximum quantity of electric charges in the sensor in this embodiment is about 150 [pC] ($Q_{\max} = 150$ [pC]). If the resistance R of the resistor 116 is determined in the aforementioned manner, an appropriate
20 value of the capacitance C of the capacitor 117 can be obtained by calculation using the expression 8.

Fig. 4 is a flow chart showing the operation

of the control circuit 12. The control circuit 12 judges whether an X-ray pulse emission start signal is received from the data transmitter-receiver circuit 3 or not. The situation of the routine goes to step 402 when
5 the X-ray pulse emission start signal is received whereas the judgment in step 401 is repeated when the X-ray pulse emission start signal is not received (step 401). When the X-ray pulse emission start signal is supplied to the control circuit 12, processing of the
10 output signals of the semiconductor sensors 21 to 2n starts after a predetermined delay time ($10\ \mu\text{s}$ in this embodiment) is passed (step 402).

First, semiconductor sensor number i to be subjected to output signal processing is set to be 1
15 (step 403). Then, a "close" command is issued to a corresponding switch 13i of the multiplexer 13 (step 404). An A-D conversion command is issued to the A-D converter 14 (step 405). When the A-D conversion by the A-D converter 14 is completed, the data is stored in the
20 memory 15 (step 406) and an "open" command is issued to the switch 13i of the multiplexer 13 (step 407). Then, a judgment is made as to whether i is equal to n or not (whether measurement on all the semiconductor sensors 21 to 2n is performed or not) (step 408). If i is equal to
25 n , the situation of the routine goes to step 409. If i is not equal to n , the situation of the routine goes to step 410. In the step 410, the calculation $i=i+1$ is made and then the situation of the routine goes back to

the step 404.

After the output signals of the semiconductor sensors 21 to 2n are A-D converted and stored in the memory 15 in the aforementioned manner, the output
5 signals stored in the memory 15 are corrected on the basis of correction coefficients stored in the correction coefficient memory 16 and are overwritten in the memory 15 again (step 409). Then, a judgment is made as to whether a data transmission request signal is
10 received from the data transmitter-receiver circuit 3 or not (step 411). At a point of time when there is the judgment that the data transmission request signal is received, the corrected output signals stored in the memory 15 are transmitted to the data transmitter-
15 receiver circuit 3 (step 412). When data transmission is completed, the situation of the routine goes back to the step 401. The output signals of the semiconductor sensors 21 to 2n are processed by the aforementioned operation of the control circuit 12.

20 Although this embodiment has shown the case where the data correction step 409 is provided between the steps 408 and 411, the invention may be applied also to the case where the data correction step is provided between the steps 406 and 407. That is, correction may
25 be made whenever the output signal of one semiconductor sensor is processed. The outputs of the integrators are, however, attenuated while data correction is made. Accordingly, higher accurate measurement can be made

when correction is made at once after all the output signals of the semiconductor sensors 21 to 2n are stored in the memory as described above in this embodiment.

As described above, according to this
5 embodiment, a dark current generated in each of the semiconductor sensors 21 to 2n can be removed because the capacitor 114 is provided for removing a DC component from the output current of each of the semiconductor sensors 21 to 2n. Furthermore, a value
10 proportional to the average number of photons in X-rays can be obtained accurately by integration by a corresponding integrator even in the case where a small number of incident photons are given because the integrator constituted by the combination of the
15 operational amplifier 115, the resistor 116 and the capacitor 117 is provided.

Although this embodiment has shown the case where the correction coefficient memory 16 is provided newly, there is no fear that the hardware amount is
20 increased by addition of the correction coefficient memory 16 because the memory 15 and the correction coefficient memory 16 can be set in one chip sufficiently. Hence, in comparison between the hardware amount in this embodiment and that in the first
25 background art, the first background art requires three IC's per semiconductor sensor for forming two operational amplifiers and one sample/hold amplifier whereas this embodiment requires one IC per

semiconductor sensor for forming one operational amplifier. In this embodiment, the number of IC's can be reduced to one thirds. If the fact that a large number of semiconductor sensors are provided is
5 considered, the hardware amount of the signal processor in this embodiment can be reduced by half as a whole compared with the first background art. According to this embodiment, the hardware amount of the signal processor can be reduced in the aforementioned manner,
10 so that the production cost can be reduced.

Incidentally, the signal processor 1 described above in this embodiment may be separated into a plurality of circuit boards so that the throughput time can be shortened by parallel processing in each of the
15 circuit boards. Fig. 8 shows a configuration in the case where five signal processors 1a to 1e are used. Incidentally, the signal processors 1a to 1e are formed on separate circuit boards, respectively. Further, as shown in Fig. 8, the bus 17 is provided in common with
20 all the signal processors 1a to 1e. In this case, the signal processors 1a to 1e can operate in parallel with one another except for data transmission to the data transmitter-receiver circuit 3. Hence, signal processing can be performed rapidly, that is, high
25 accurate measurement can be made before the output signals of the first-stage circuits are attenuated. Incidentally, because data transmission to the data transmitter-receiver circuit 3 is made in accordance

with the request of the data transmitter-receiver circuit 3, there is no risk of crosstalk. Further, the number of circuit boards can be increased arbitrarily if addresses are allocated to the signal processors in advance.

Although the aforementioned embodiment has shown the case where the output signals of the first-stage circuits 111 to 11n are processed successively at intervals of 5 μ s, the processing interval may be shortened if the time required for the A-D conversion in the A-D converter 14 can be shortened. Further, the control circuit 12 may be produced in combination with a digital circuit or may use a one-chip microcomputer. Although circuits for supplying power to the respective electronic circuits are not shown, it is a matter of course that such power supply circuits are connected to the respective electronic circuits in the first-stage circuits 111 to 11n.

(Embodiment 2)

Referring to Fig. 5, a signal processor as another embodiment of the present invention will be described below. This embodiment is different from the embodiment 1 in that a data collection IC in which sample/hold circuits and a multiplexer are incorporated in an A-D converter is used in this embodiment. The point of difference from the embodiment 1 will be described below.

As shown in Fig. 5, the data collection IC 50

has sample/hold circuits 51 to 5n, multiplexing switches 131 to 13n, an A-D converter 14, and a memory 15. The output signals of the first-stage circuits 111 to 11n are supplied to the sample/hold circuits 51 to 5n in the data collection IC and held immediately by the sample/hold circuits 51 to 5n respectively. The held output signals are supplied to the A-D converter through the switches 131 to 13n respectively. After A-D conversion, the output signals are stored in the memory 15. Incidentally, the same method as in the embodiment 1 is used as a method for controlling the switches 131 to 13n, the A-D converter 14 and the memory 15.

Also in this embodiment, like the embodiment 1, a dark current generated in each of the semiconductor sensors 21 to 2n can be removed by a capacitor 114 and a value proportional to the average number of photons in X-rays can be obtained accurately by integration in an integrator even in the case where a small number of incident photons are given. Furthermore, because the output signals just after X-ray irradiation are held by the sample/hold circuits 51 to 5n respectively, correction of the output signals as described above in the embodiment 1 is not required. Hence, the correction coefficient memory 16 is not required. Alternatively, the sample/hold circuits 51 to 5n may be provided on the first-stage circuits 111 to 11n.

(Embodiment 3)

Referring to Fig. 6, a signal processor as a

further embodiment of the present invention will be described below. This embodiment is different from the embodiment 2 in that logarithmic transformation circuits are added to the data collection IC. The point of
5 difference from the embodiment 2 will be described below.

In the data collection IC 60, the output signals held by the sample/hold circuits 51 to 5n are supplied to the logarithmic transformation circuits 61
10 to 6n respectively. Upon reception of the output signals, the logarithmic transformation circuits 61 to 6n transform the output signals into logarithmic values and output the logarithmic values respectively. In the CT controller 9, the data required in calculation for
15 reconstructing a perspective image are logarithmic-value data. Hence, because logarithmic transformation is performed in the signal processor 1 in advance, logarithmic transformation needs not to be performed in the CT controller 9. Calculation in the CT controller 9
20 can be simplified. Although this embodiment has shown the case where the logarithmic transformation circuits 61 to 6n are provided for performing logarithmic transformation, such calculation may be made by a software means in the control circuit 12 before data
25 transmission.

(Embodiment 4)

Referring to Fig. 7, a signal processor as a further embodiment of the present invention will be

described below. Incidentally, the point of difference of this embodiment from the embodiment 1 will be mainly described below.

In a first-stage circuit 70 in Fig. 7,
5 switches 71 and 72 are connected to a junction between the semiconductor sensor 21 and the resistor 119 so as to be parallel with each other. A switch 73 is further connected to the other end of the switch 71. A
10 capacitor 74 is connected to a junction between the switches 71 and 73 while the other end of the capacitor 74 is connected to the ground. On the other hand, a switch 75 is further connected to the other end of the switch 72. A capacitor 76 is connected to a junction between the switches 72 and 75 while the other end of
15 the capacitor 76 is connected to the ground.

The operation of the first-stage circuit 70 will be described. Before X-ray pulse irradiation starts, the switches 71 and 72 are closed and the switches 73 and 75 are opened. At the same time when X-
20 ray pulse irradiation starts, the switch 71 is opened. On the other hand, the switch 72 is kept close during the X-ray pulse irradiation and the switches 73 and 75 are kept open during the X-ray pulse irradiation. Because the switch 72 is kept close during the X-ray
25 pulse irradiation, electric charges due to a current, inclusive of a dark current, generated in the semiconductor sensor 21 by incidence of X-rays are stored (integrated) in the capacitor 76. Because the

switch 71 is closed after the completion of the X-ray pulse irradiation, electric charges due to the dark current are stored (integrated) in the capacitor 74. Incidentally, in this embodiment, the resistance of the resistor 119 is set to be as large as 1 [M Ω] and the capacitance of the capacitors 74 and 76 is set to be equal to the capacitance 100 [pF] of the semiconductor sensor 21. As a result, the current generated by the X-ray pulse is prevented from flowing into the resistor 119 so that the current is made to flow into the semiconductor sensor 21 and the capacitor 76 equally.

Then, the switches 73 and 75 are closed, so that currents corresponding to the electric charges stored in the capacitors 74 and 76 flow into posterior circuits through the switches 73 and 75 respectively. An amplifier unit constituted by a combination of an operational amplifier 81, a resistor 82 and a capacitor 83 is connected to the switch 73. An amplifier unit constituted by a combination of an operational amplifier 84, a resistor 85 and a capacitor 86 is connected to the switch 75. The two amplifier units amplify input signals by the same amplification factors respectively and supply the amplified signals to a subtractor 87. Upon reception of the output signals of the two amplifier units, the subtractor 87 subtracts one signal from the other and supplies a result of the subtraction to an sample/hold circuit 88. That is, the subtractor 87 calculates the difference between a voltage

corresponding to a current, inclusive of a dark current,
generated in the semiconductor sensor 21 by incidence of
X-rays and a voltage corresponding to the dark current.
Hence, the output of the subtractor 87 is provided as a
5 voltage corresponding to the current generated in the
semiconductor sensor 21 by incidence of X-rays. Upon
reception of the output signal of the subtractor 87, the
sample/hold circuit 88 holds the signal and supplies the
signal to the A-D converter 14. Upon reception of the
10 output signal of the sample/hold circuit 88, the A-D
converter 14 A-D converts the signal.

According to this embodiment, a voltage
corresponding to a current generated by incidence of X-
rays can be calculated accurately because the subtractor
15 87 calculates the difference between a voltage
corresponding to a current inclusive of a dark current
generated in each of the semiconductor sensors 21 to 2n
by incidence of X-rays and a voltage corresponding to
the dark current so that the voltage corresponding to
20 the dark current generated in each of the semiconductor
sensors 21 to 2n is removed. This technique is
effective regardless of the number of incident photons.

Incidentally, the switch 71, the capacitor 74,
the switch 73 and the amplifier unit using the
25 operational amplifier 81 can be omitted. In this case,
a voltage corresponding to the dark current is measured
without X-ray irradiation and stored in the memory 15
before the measurement of the object starts. When the

stored voltage corresponding to the dark current is subtracted from the measured value at the time of the measurement of the object, a voltage corresponding to the current generated by incidence of X-rays can be
5 calculated accurately.

Although the aforementioned embodiments have shown the case where capacitors are used for removing dark currents, the invention may be applied also to the case where the capacitors are replaced by filters so
10 long as the filters can remove DC components. Although the aforementioned embodiments have shown the case where semiconductor sensors are used as X-ray sensors because X-ray CT systems using high-energy X-ray pulses are provided as subjects of the embodiments, the invention
15 may be applied also to the case where photodiodes, or the like, for measuring scintillator light or fluorescence are used as the sensors. In addition, the signal processors in the embodiments may be applied to other systems than the X-ray CT systems so long as the
20 systems can emit pulse-like X-rays and measure the X-rays.



WE CLAIM:

1. An X-ray sensor signal processor for processing an output signal of an X-ray sensor for detecting pulse-like X-rays emitted from an accelerator and passed through an inspection object, characterized in that

said X-ray sensor signal processor comprises a filter for removing a DC component from said output signal of said X-ray sensor, and an integrator for integrating said output signal of said X-ray sensor after removal of said DC component by said filter.

2. An X-ray sensor signal processor according to Claim 1, characterized in that:

said integrator outputs an output signal in accordance with a time constant; and

said time constant is set to a value by which said output signal of said integrator is made zero in a period between emission of an X-ray pulse from said accelerator and emission of the next X-ray pulse from said accelerator.

3. An X-ray sensor signal processor according to Claim 2, characterized by further comprising an A-D converter for A-D converting said output signal of said integrator, and means for correcting said output signal of said integrator obtained by A-D conversion by said A-D converter on the basis of a time necessary for said A-D converter to perform the A-D conversion after emission of said output signal from said integrator and a time

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constant of said integrator.

4. An X-ray sensor signal processor according to Claim 2, characterized by further comprising a hold circuit for holding said output signal of said integrator.

5. An X-ray sensor signal processor according to Claim 4, characterized by further comprising a logarithmic transformation circuit for transforming said output signal of said integrator held by said hold circuit into a logarithmic value.

6. An X-ray sensor signal processor for processing an output signal of an X-ray sensor for detecting pulse-like X-rays emitted from an accelerator and passed through an inspection object, characterized in that

said X-ray sensor signal processor comprises a first resistor with one end connected to said X-ray sensor and the other end connected to the ground, a first capacitor with one end connected to a junction point between said X-ray sensor and said first resistor, an operational amplifier with an inversion input connected to the other end of said first capacitor, and a combination of a second resistor and a second capacitor connected in parallel to said operational amplifier.

7. An X-ray sensor signal processor according to Claim 6, characterized by further comprising an A-D converter for A-D converting an output signal of said

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operational amplifier, wherein capacitance C of said second capacitor is set to be in a range represented by an expression:

$$Q_{\max}/V_{fs} \leq C \leq \min\{(Q_{\max} \times 10^{-m} \times 2^n)/V_{fs}, T_p/(R(n+1 \ln 2))\}$$

in which R is a resistance of said second resistor, $Q_{\max}$ is a maximum quantity of electric charges generated by said X-ray sensor, m is a dynamic range required for measurement of X-rays, V_{fs} is a maximum voltage input to said A-D converter, n is a required resolution, and T_p is a time between an emission of X-ray and an emission of next X-ray.

8. An X-ray sensor signal processor according to Claim 7, characterized by further comprising means for calculating a corrected output signal V_0 in accordance with an expression:

$$V_0 = V \times \exp(t/\tau)$$

in which V is the output signal of said operational amplifier obtained by A-D conversion by said A-D converter, t is a time between completion of emission of X-rays and start of the A-D conversion by said A-D converter, and τ is a time constant obtained as a product of the resistance R and the capacitance C.

9. An X-ray sensor signal processor for processing output signals of a plurality of X-ray

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sensors for detecting pulse-like X-rays emitted from an accelerator and passed through an inspection object, characterized in that

said X-ray sensor signal processor comprises: a plurality of first-stage circuits provided to correspond to said plurality of X-ray sensors respectively, each of said first-stage circuit including a filter for removing a DC component from an output signal of a corresponding X-ray sensor, and an integrator for integrating the output signal of said corresponding X-ray sensor from which the DC component is removed by said filter; an A-D converter for A-D converting output signals of said plurality of first-stage circuits successively; and means for correcting said output signals of said first-stage circuits obtained by the A-D conversion by said A-D converter on the basis of a time between emission of an output signal from each of said integrators and completion of the A-D conversion by said A-D converter and a time constant of each of said integrators after all the output signals of the plurality of first-stage circuits are A-D converted by said A-D converter.

10. An X-ray sensor signal processor for processing an output signal of an X-ray sensor for detecting pulse-like X-rays emitted from an accelerator and passed through an inspection object, characterized in that

said X-ray sensor signal processor comprises a

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first capacitor for integrating the output signal of said X-ray sensor when X-rays are not detected by said X-ray sensor, a second capacitor for integrating the output signal of said X-ray sensor when X-rays are being detected by said X-ray sensor, and a subtractor for subtracting the output signal of said X-ray sensor integrated by said first capacitor from the output signal of said X-ray sensor integrated by said second capacitor.

11. An X-ray computed tomography system comprising an accelerator for emitting pulse-like X-rays, a scanner for rotating or moving up and down an inspection object irradiated with X-rays emitted from said accelerator, a plurality of X-ray sensors for detecting X-rays passed through said object irradiated with said X-rays, an X-ray sensor signal processor for processing output signals of said plurality of X-ray sensors, and a computed tomography controller for reconstructing a perspective image of a section of said object on the basis of the output signals of said X-ray sensors processed by said X-ray sensor signal processor, characterized in that

said X-ray sensor signal processor includes filters for removing DC components from the output signals of said X-ray sensors respectively, and integrators for integrating the output signals of said X-ray sensors respectively after removal of the DC components by said filters.



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12. An X-ray computed tomography system according to Claim 11, characterized in that:

each of said integrators outputs an output signal in accordance with a time constant; and

said time constant is set to a value by which the output signal of a corresponding integrator reaches zero in a time between an emission of an X-ray from said accelerator and an emission of next X-ray from said accelerator.

13. An X-ray computed tomography system according to Claim 12, characterized by further comprising an A-D converter for A-D converting the output signals of said integrators, and means for correcting the output signals of said integrators obtained by A-D conversion by said A-D converter on the basis of a time between emission of an output signal from each of said integrators and completion of the A-D conversion by said A-D converter and time constants of said integrators.

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