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(54) **STATIC CHARGE FILTER FOR CARDIAC ELECTRICAL SIGNALS CORRUPTED WITH STATIC CHARGE AND ROUTED TO AN ELECTROCARDIOGRAPH (EKG) MONITOR**

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(57)

ABSTRACT

A static charge filter (522) removes static charge in a cardiac electrical signal. The static charge filter includes a first amplifier (608) configured to amplify an input signal, which includes the cardiac electrical signal and static charge from an electrode, which is in a path of an X-ray beam. The static charge filter further includes a limiter (614) configured to limit a maximum voltage of the signal based on a predetermined clamping threshold, producing a voltage clamped signal. The static charge filter further includes a filter (616) configured to filter high frequency components of the voltage clamped signal, producing a filtered signal. The static charge filter further includes a second amplifier (620) configured to scale an amplitude of the filtered signal so that cardiac electrical signal in an output signal has a same voltage level as a voltage level of the cardiac electrical signal in the input signal.

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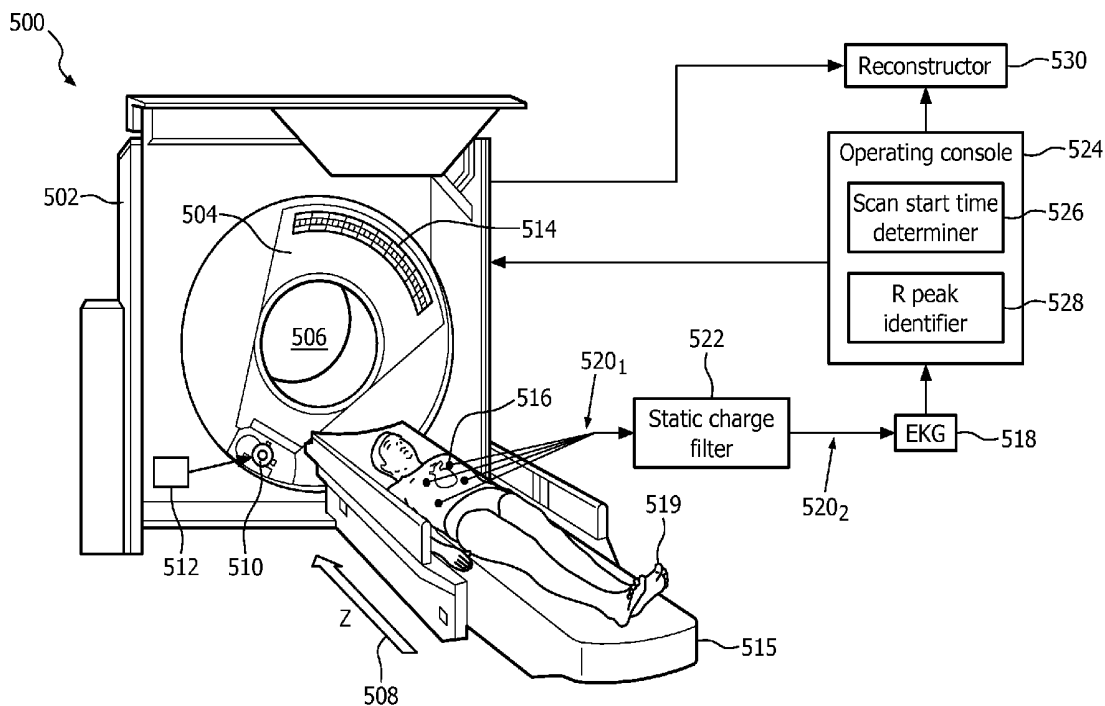
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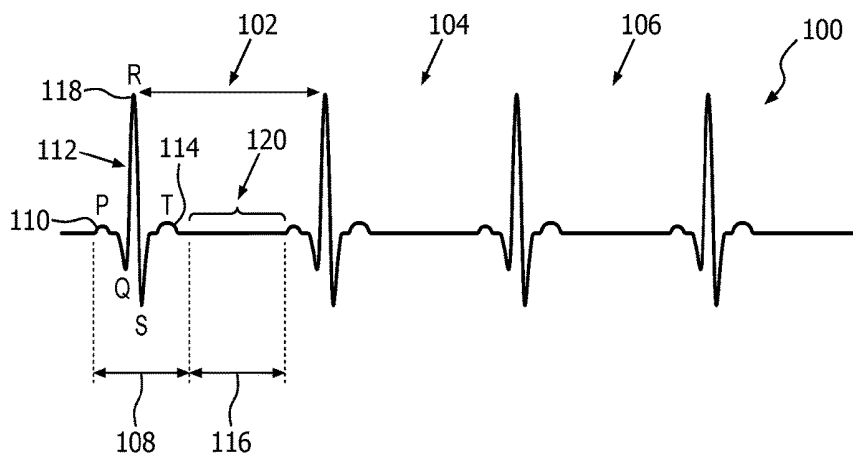


FIG. 1

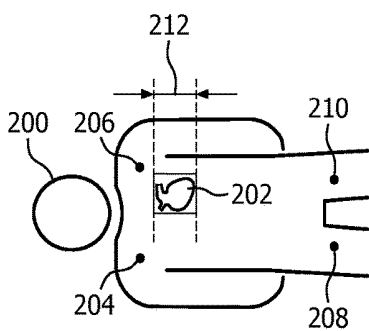


FIG. 2

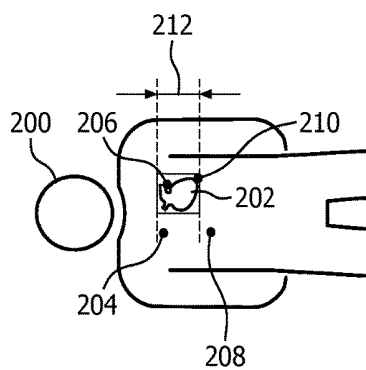


FIG. 3

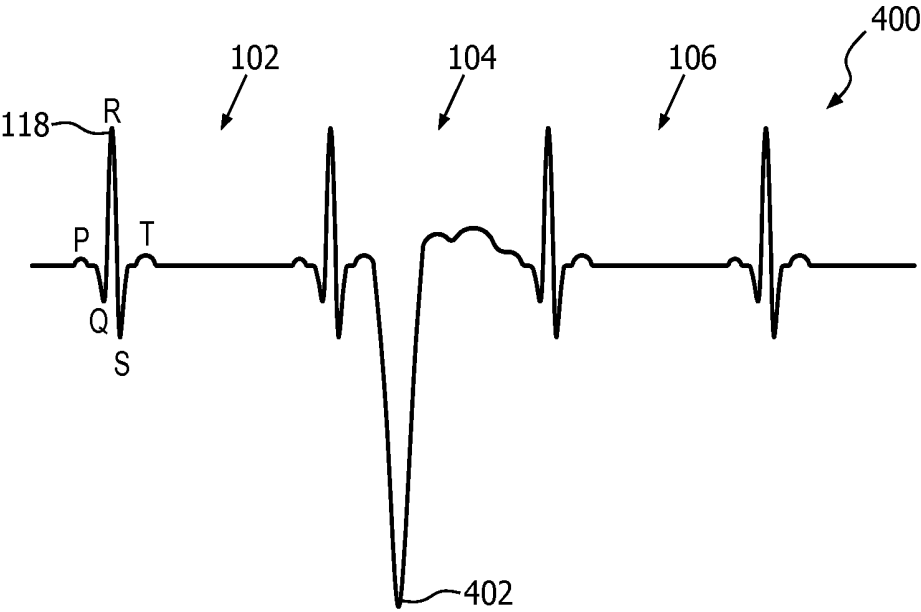


FIG. 4

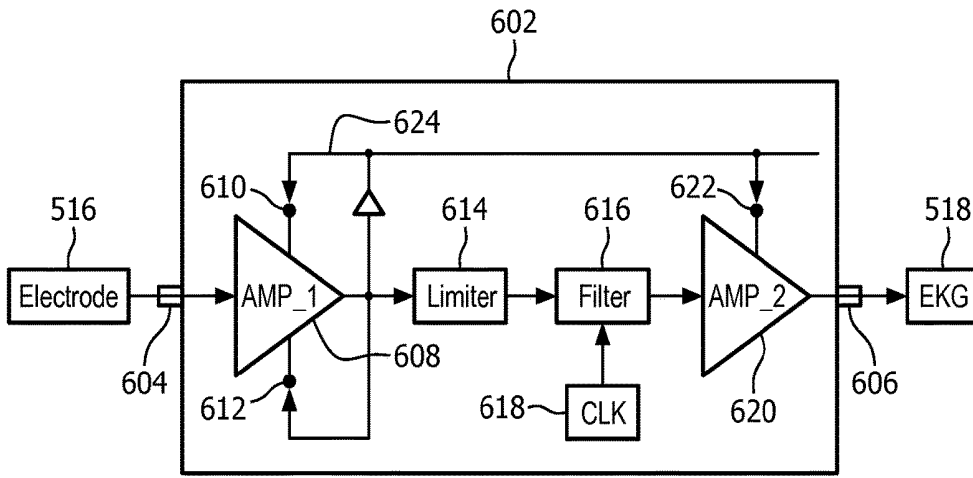


FIG. 6

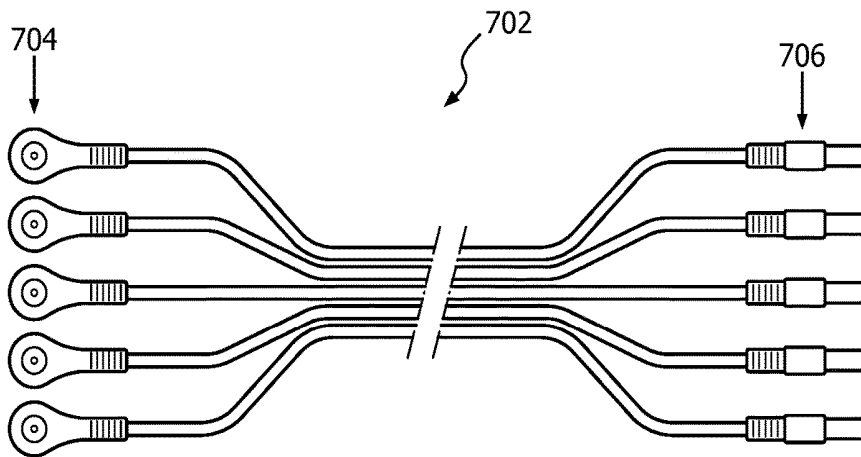


FIG. 7

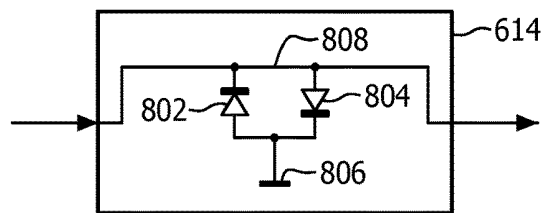


FIG. 8

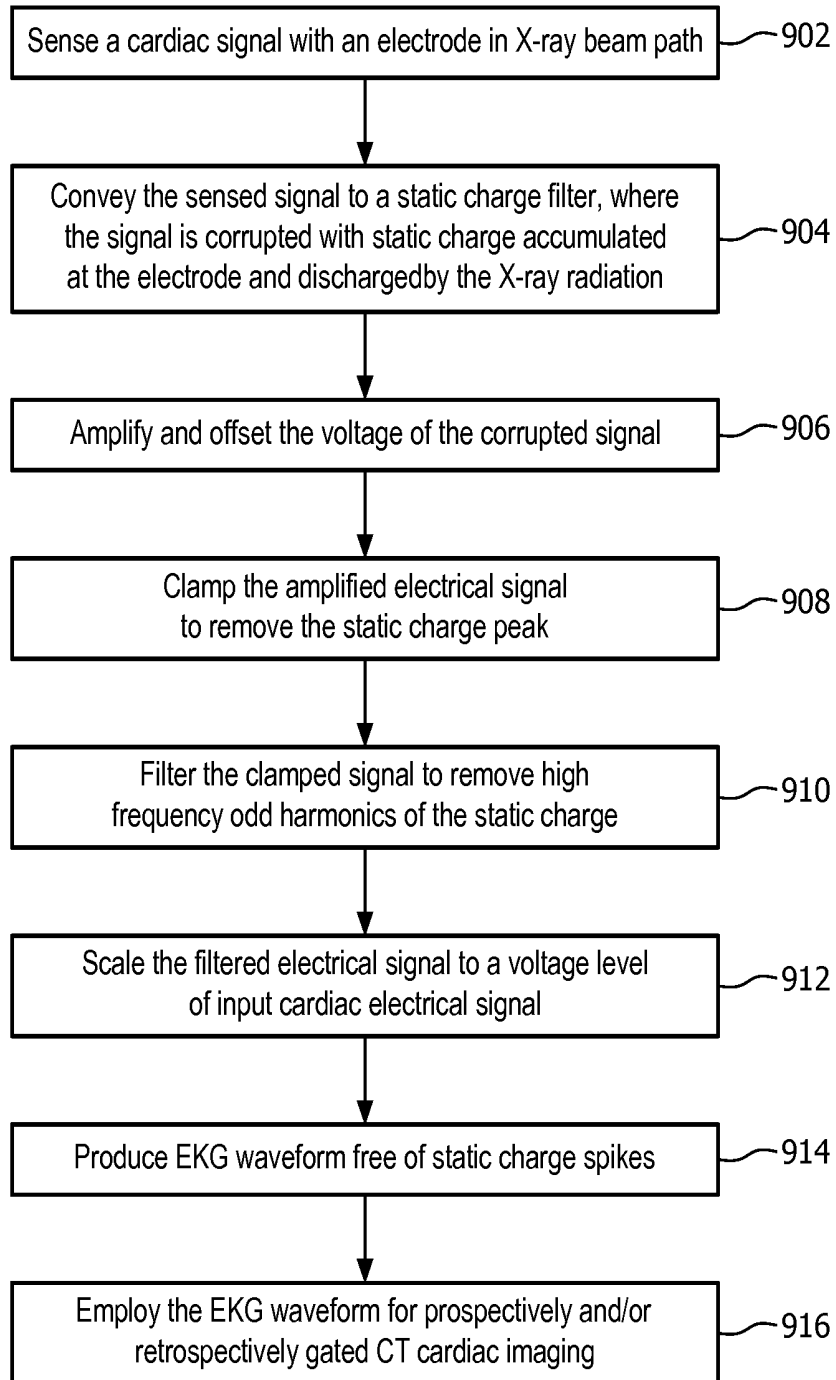


FIG. 9

STATIC CHARGE FILTER FOR CARDIAC ELECTRICAL SIGNALS CORRUPTED WITH STATIC CHARGE AND ROUTED TO AN ELECTROCARDIOGRAPH (EKG) MONITOR

FIELD OF THE INVENTION

[0001] The following generally relates to a static charge filter for cardiac electrical signals corrupted with static charge and routed to an electrocardiograph (EKG) monitor, and is described with particular application to computed tomography (CT), where EKG electrodes and/or leads in a path of an X-ray radiation beam accumulate charge, which is discharged by the X-ray radiation through the leads, corrupting the cardiac electrical signals routed through the leads to the EKG monitor.

BACKGROUND OF THE INVENTION

[0002] Cardiac gated Computed tomography (CT) generally requires a CT scanner, an electrocardiograph (EKG) monitor, patient electrodes, and leads that mechanically connect to and that electrically connect the electrodes and the EKG monitor. The EKG electrodes sense electrical changes on the skin that arise from the heart muscle depolarizing during each heartbeat. The electrical charge is transferred via the EKG electrodes to the leads, which routes the electrical charge to the EKG monitor, which processes the charge and produces an EKG waveform indicative of electrical activity of each heart beat or cycle, which reflects a state of the heart throughout the heart cycle. FIG. 1 shows an example of a "normal" waveform **100** for three cardiac cycles **102**, **104** and **106**, where a y-axis represents amplitude (e.g., in millivolts) and an x-axis represents time.

[0003] In FIG. 1, each of the cardiac cycles **102-106** has a systolic period **108** in which the atria (P wave **110**) and subsequently the ventricles (QRS complex **112**) contract and the ventricles then re-polarize (the T wave **114**), and a diastolic period **116** in which the heart relaxes and refills with circulating blood. For prospectively gated cardiac CT, an R peak **118** of the QRS complex **112** is identified in the waveform **100**, e.g., based on its amplitude, and used to trigger a scan for a predetermined acquisition window within a quiet phase **120** of the heart in which motion of the heart is relative low compared to the motion during other phases of heart. For retrospectively gated cardiac CT, the heart is scanned and the projection data is synchronized with the waveform **100** so that after the scan the R peak **118** can be identified and used to retrieve projection data for the acquisition window. In both instances, multiple images for the window but different cycles can be reconstructed.

[0004] The number and placement of the electrodes on the patient depends on whether a 3, 4, 5 or 12-lead configuration is used. FIG. 2 show an example of EKG electrode placement on a patient **200** relative to the heart **202** for a 4-lead configuration. The illustrated configuration includes a right arm electrode **204**, a left arm electrode **206**, a right leg electrode **208** and a left leg electrode **210**. With this configuration, the entire heart **202** can be scanned without any of the electrodes **204-210** in a path **212** of the X-ray beam. FIG. 3 shows an example where all of the electrodes but one (the electrode **208**) are in the path **212** of the X-ray beam. Unfortunately, electrodes and/or leads in the X-ray beam accumulate static charge, and the X-ray radiation discharges the charge (via electrostatic discharge, or ESD), which

results in a voltage spike or pulse that ultimately corrupts the EKG waveform. FIG. 4 shows an example of a corrupted waveform.

[0005] The waveform **400** of FIG. 4 is substantially the same as the waveform of FIG. 1, except the waveform **400** include a voltage spike **402**, which has an amplitude at least a large as that of the R peak **118**. In this example, the amplitude of the spike **402** is approximately two and a half times the amplitude of the R peak **118** and in the opposite direction (i.e., negative). However, such spikes may be larger or smaller, and/or positive. Unfortunately, the spike **402** may be identified as an R peak. With prospectively gated cardiac CT, a false R peak may trigger a scan that is at least partially outside of the acquisition window within the quiet phase. With retrospectively gated cardiac CT, the false R peak may lead to reconstructing projection data acquired at least partially outside of the acquisition window within the quiet phase. In either instance, this may result in degradation of image quality (increased motion), possibly to the extent that the final image is not diagnostic quality, even though the patient has been irradiated.

[0006] One approach to mitigating such static charge is to keep the electrodes and leads out of the beam path. Unfortunately, this is not always feasible, e.g., where the electrodes are already on a patient in an emergency situation. Another approach includes applying a contact gel to the electrodes before scanning to provide a discharge path. Likewise, this approach is not always feasible. Furthermore, it requires access to contact gel and adds additionally scan preparation steps and cost. Another approach is to use carbon impregnated electrodes and leads which bleed off charge before it is large enough to corrupt the cardiac electrical signals. Such electrodes are described in US 20140316231 A1 to Luhta et al. Unfortunately, such electrodes are customized and may add cost, which may deter a facility from procuring them, especially with the small percentage of cases that require them.

SUMMARY OF THE INVENTION

[0007] Aspects described herein address the above-referenced problems and/or others.

[0008] In one aspect, a static charge filter removes static charge in a cardiac electrical signal. The static charge filter includes a first amplifier configured to amplify an input signal, which includes the cardiac electrical signal and static charge from an electrode, which is in a path of an X-ray beam. The static charge filter further includes a limiter configured to limit a maximum voltage of the signal based on a predetermined clamping threshold, producing a voltage clamped signal. The static charge filter further includes a filter configured to filter high frequency components of the voltage clamped signal, producing a filtered signal. The static charge filter further includes a second amplifier configured to scale an amplitude of the filtered signal so that cardiac electrical signal in an output signal has a same voltage level as a voltage level of the cardiac electrical signal in the input signal.

[0009] In another aspect, a system includes a gantry with an examination region, an X-ray radiation source configured to produce and transmit X-ray radiation that traverses the examination region, an X-ray sensitive detector array configured to detect X-ray radiation traversing the examination region and generate projection data, and a reconstructor configured to reconstruct at least a sub-portion of the pro-

jection data and generate an image indicative thereof. The system further includes a static charge filter configured to receive cardiac electrical signals sensed by electrodes in the examination region, wherein at least one of the electrodes is in a path of the X-ray radiation traversing the examination region and remove static charge from the cardiac electrical signal.

[0010] In another aspect, a method includes receiving a cardiac electrical signal from an electrode in a path of an X-ray beam during an imaging examination, wherein the cardiac electrical signal includes static charge. The method further includes removing a static charge voltage spike from the received cardiac electrical signal, producing a static charge filtered cardiac electrical signal. The method further includes transmitting the static charge filtered cardiac electrical signal to an EKG monitor, which processes the static charge filtered cardiac electrical signal and produces an EKG waveform representing an electrical activity of a heart of the subject.

BRIEF DESCRIPTION OF THE DRAWINGS

[0011] The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for purposes of illustrating the preferred embodiments and are not to be construed as limiting the invention.

[0012] FIG. 1 schematically illustrates an example “normal” a EKG waveform.

[0013] FIG. 2 schematically illustrates example placement of EKG electrodes on a patient outside of the X-ray beam path.

[0014] FIG. 3 schematically illustrates example placement of the EKG electrodes on a patient with at least one electrode in the X-ray beam path.

[0015] FIG. 4 schematically illustrates an example of a EKG waveform corrupted with static charge from an EKG electrode and/or lead in the X-ray beam path.

[0016] FIG. 5 schematically illustrates an example system including an imaging scanner, and EKG device, and a static charge filter.

[0017] FIG. 6 schematically illustrates an example of the static charge filter.

[0018] FIG. 7 shows an example set of EKG leads.

[0019] FIG. 8 schematically illustrates an example of a voltage limiter of the static charge filter.

[0020] FIG. 9 illustrates an example method for removing static charge corrupting an EKG waveform, where the static charge is a results of X-ray induced discharge of static charge accumulated on an EKG electrode and/or lead in the X-ray beam path.

DETAILED DESCRIPTION OF EMBODIMENTS

[0021] FIG. 5 illustrates an imaging system 500 such as a computed tomography (CT) scanner. The illustrate system 500 includes a generally stationary gantry 502 and a rotating gantry 504. The rotating gantry 504 is rotatably supported by the stationary gantry 502 via a bearing or the like and rotates around an examination region 506 about a longitudinal or z-axis 508 and emits radiation.

[0022] A radiation source 510, such as an x-ray tube, is supported by the rotating gantry portion 504 and rotates therewith. The radiation source 510, as it rotates around the examination region 506, emits radiation that traverse the

examination region 506. The radiation source 510 can also emit radiation while the rotating gantry 504 is at a static position, for example, for a pilot, scout, axial, and/or other scan.

[0023] A radiation source controller 512 selectively turns radiation on and off. For example, the radiation source controller 512 can “gate” the radiation source 510, based on a gating signal, to selectively turn x-rays on to acquire data only in an acquisition window. For example, the gating signal can be generated in response to detecting an R peak in an EKG signal and cause the imaging system 500 scan the heart during a predetermined acquisition window during a quiet phase of the heart for a gate cardiac CT scan.

[0024] An X-ray radiation sensitive detector array 514 subtends an angular arc opposite the examination region 506 relative to the radiation source 510. The illustrated X-ray radiation sensitive detector array 514 includes a one or two-dimensional array of photosensitive pixels. The X-ray radiation sensitive detector array 514 detects The X-ray radiation traversing the examination region 506 and generates projection data, or a signal, indicative thereof.

[0025] A subject support 515, such as a couch, supports a subject, such as a human or animal, or an object within the examination region 506. The subject support 515 is movable horizontally and/or vertically, which enables an operator or the system to load a subject, suitably position the subject within the examination region 506 before, during and/or after scanning, and unload the subject.

[0026] EKG electrodes 516 are attached to a patient 519 and sense electrical signals of the heart (cardiac electrical signals). A first set of leads 520₁ routes the cardiac electrical signals from the EKG electrodes 516. Where static charge is present and discharged from the EKG electrodes 516 and/or the first set of leads 520₁ by the X-ray beam and/or otherwise, the static charge corrupts the cardiac electrical signals (e.g., with one or more spikes 402 shown in FIG. 4), and the first set of leads 520₁ routes the corrupted cardiac electrical signal, which includes the cardiac electrical signals and static charge spikes.

[0027] A static charge filter 522 is connected to the other end of the first set of leads 520₁, receives the corrupted cardiac electrical signals, filters the corrupted cardiac electrical signal to remove static charge, and outputs filtered cardiac electrical signals, which are essentially free of static charge that may result in a false R peak. A second set of leads 520₂ routes the filtered cardiac electrical signals from the static charge filter 522. An EKG monitor 518 is connected to the other end of the second set of leads 520₂, and processes the filtered cardiac electrical signals and generates an EKG waveform similar to the waveform 100 shown and described in connection with FIG. 1.

[0028] As described in greater detail below, in one instance the static charge filter 522 includes active circuitry that amplifies an input corrupted cardiac electrical signal, clamps a maximum allowable voltage amplitude of the amplified signal, low pass filters high frequency components resulting from the clamping, and scales the voltage amplitude of the filtered signal to a level of the input cardiac electrical signals, to remove static charge. Thus, with prospectively and/or retrospectively gated cardiac CT, a voltage spike due to static charge will not be identified in the signal output by the static charge filter 522 as an R peak and used as a trigger. As such, the image quality of the reconstructed images will not be degraded due to static charge when an

electrode(s) of the electrodes **516** and/or a lead(s) of the first set of leads **520₁** is in the path of the X-ray beam during scanning.

[0029] A computing system serves as an operator console **524** and allows a user to select an imaging protocol such as a prospectively and/or retrospectively gated cardiac CT protocol. The illustrated operator console **524** includes an R peak identifier **528** and a scan start time determiner **526**. The R peak identifier **528** identifies R peaks in the EKG waveform and generates a trigger signal indicative thereof for both prospectively and retrospectively gated scans. For prospectively gated scans, a scan start time determiner **526** determines a start time for an acquisition window based on the trigger signal and/or other information such as a time delay from the R peak. In a variation, the R peak identifier **528** is located in the EKG **518** and/or other device. Where the system is configured only for retrospectively gated cardiac CT scans, the scan start time determiner **526** can be omitted.

[0030] A reconstructor **530** reconstructs the projection data and generates volumetric image data indicative thereof. For prospectively gated cardiac CT, this includes reconstructing the projection data for the scan of the acquisition window within the quiet phase triggered by the R peak of the EKG waveform from the filtered cardiac electrical signals. For retrospectively gated cardiac CT, this includes identifying projection data corresponding to the acquisition window within the quiet phase based on the filtered cardiac electrical signals, which is synchronized with the projection data from the scan. The resulting volumetric image data can be visually presented via a display monitor, stored in a data repository (e.g., a picture and archiving communication system, or PACS), etc.

[0031] FIG. 6 schematically illustrates an example channel **602** of the static charge filter **522**.

[0032] In one instance, the static charge filter **522** includes a channel for each EKG electrode. For example, the static charge filter **522** for a 4 lead EKG will include at least 4 channels, etc. In one configuration, each channel filters a corresponding one of the input cardiac electrical signals. This configuration is well-suited for EKGs which can vary which set of the electrical signals is used to identify the R peak. In another configuration, where a signal from a predetermined particular electrode is used to identify the R peak, only the channel corresponding to the particular electrode includes the filter, and the other channels are pass-throughs.

[0033] The illustrated channel **602** includes an input connector **604** and an output connector **606**. In one instance, the input connector **604** is configured the same as an input connector of the EKG monitor **518**, and the output connector **606** is configured the same as connector of the electrode **516**. In this manner, a same cable utilized without the static charge filter **522** to electrically connect the EKG electrodes **516** to the EKG **518** can be used to connect the EKG electrodes **516** to the static charge filter **522** and the static charge filter **522** to the EKG **518**. FIG. 7 shows an example cable **702** with electrode connectors **704** and EKG connectors **706** for a 5-lead EKG. Other types of connectors (e.g., BNC, etc.) are also contemplated herein.

[0034] The illustrated channel **602** further includes a first amplifier **608** that amplifies the input signal. The first amplifier **608** includes a first gain input **610** and an offset input **612**. A limiter **614** is configured to clip or limit the

amplified voltage of the amplified signal. A clocked filter **616** and a clock **618** filters the clipped signal. A second amplifier **620** with a second gain input **622** scales the filtered signal. A first gain of the first amplifier **608** is set to ensure the cardiac electrical signal is amplified near a clipping threshold of the limiter **614** so that spikes **402** (FIG. 4) due to X-ray static discharge events are clamped by the limiter **614**, which generates out of band harmonics. Automatic gain control is achieved through a feedback loop **624**. The offset input is configured to monitor the root-mean-square (RMS) and/or other statistical measure and keep the offset voltage close to zero.

[0035] FIG. 8 schematically illustrates an example of a suitable limiter **614**. In this example, the limiter **614** includes two diodes **802** and **804** in parallel (in reverse polarity) with one of the sides connected to electrical ground **806** and the other of the sides connected to an electrically conductive pathway **808** through the limiter **614**. With this configuration, as long as the amplitude of the input signal does not exceed the reverse bias voltage of the diodes, the diodes remain off and the signal passes through. However, when the amplitude of the signal exceeds the reverse bias voltage of one of the diodes, that diode turns on and shorts to the electrical ground **806**. As such, the limiter **614** clamps the signal such that the output signal never exceeds the reverse bias voltage. Other voltage limiting circuitry is also contemplated herein.

[0036] Returning to FIG. 6, a suitable filter **616** is a clock programmable (switched capacitor) low pass filter. In this example, the filter **616** is a high order low pass type filter that rejects higher frequency content, or the out of band harmonics generated after the limiter **614** clamps the signal. The filtered signal is substantially free of the static discharge seen by the electrodes in the X-ray beam. The second amplifier **620** is a low output impedance amplifier configured with a second gain that scales the processed signal to match the levels of the input electrical signal. In one instance, where the first gain of the first amplifier **608** is A , the second gain of the second amplifier **620** is $1/A$. In another instance, the second gain of the second amplifier **620** is $1/A + \text{internal losses}$.

[0037] The circuitry implementation of the channel **602** can be realized in a standard CMOS fab processing steps of an ASIC, Commercial-Off-The-Shelf (COTS) components, and/or otherwise. COTS components may include power supply regulators, low-offset operational amplifiers (Op-Amps), clock-programmable filters, limiter circuits including simple versions with diode clamps, and standard discrete capacitors and resistors with printed circuit (PC) card and standard connectors and wiring. Power can be supplied by the EKG **518**, the imaging system **500**, an internal (rechargeable or non-rechargeable) battery, AC power from a wall receptacle, AC and/or DC power from a power supply, and/or other power source.

[0038] The following provides a non-limiting example.

[0039] For this example, the amplitude of the R peaks **118** (FIG. 1) in the cardiac electrical signals from the electrodes **204-210** (FIG. 2) is in a millivolt (mV) range. The two diodes **802** and **804** (FIG. 8) have a reverse bias voltage of 0.7V (0.3V for germanium diodes). The first gain is $A=1000$, and the second gain is $1/1000$. In this instance, the voltage of the amplified signal will be clamped at $\pm 0.7V$, and the spike **402** will be removed regardless of whether it is positive or negative (e.g., as shown in FIG. 4). After low pass filtering

the high frequency odd harmonics, the filtered signal is scaled by the second gain, and the output signal will be in the millivolt range like the input signal. Where there is no spike 402 (FIG. 4), the channel 602 essentially behaves as a pass through.

[0040] FIG. 9 illustrates an example method for removing static charge corrupting an EKG waveform, where the static charge is a results of X-ray induced discharge of static charged accumulated on an EKG electrode and/or lead in the X-ray beam path.

[0041] At 902, cardiac electrical signals are sensed by EKG electrodes, where at least one of the electrodes is in a path of an X-ray beam, and the at least one of the electrodes has accumulated static charged that is discharged by the X-ray radiation incident on the at least one of the electrodes through at least one of the leads. Additionally or alternatively, at least one of the leads in the path of the X-ray beam has accumulated static charge that is discharged by the X-ray radiation incident on the at least one of the leads.

[0042] At 904, the cardiac electrical signals and the discharged static charge are electrically routed to a static charge filter, wherein the static charge filter processes one or more of the corrupted cardiac electrical signals from the different electrodes through separate channels. The corrupted cardiac electrical are conveyed through conventional EKG cables such as those shown in FIG. 3 and/or otherwise.

[0043] At 906, the corrupted cardiac electrical signal received by the static charge filter, which includes both the cardiac electrical signals and the discharged static charge, is amplified and offset. As discussed herein, the amplification is based on the voltage clamping level of the limiter and the offset keeps the RMS value close to zero (e.g., ± 10 mV).

[0044] At 908, the amplified and offset signal is clamped based on the clamping threshold of a limiter, which removes static charge spike(s).

[0045] At 910, high frequency odd harmonics resulting from the clamping are filtered by a low pass filter.

[0046] At 912, the filtered signal is scaled by a second amplifier so that the output cardiac electrical signals have an amplitude corresponding to the amplitude of the input cardiac electrical signals.

[0047] At 914, the processed cardiac electrical signals are conveyed to an EKG monitor, which processes these signals and produces a waveform such as the EKG waveform 100 (FIG. 1), which will be free of the static charge spike 402 (FIG. 4).

[0048] At 916, one or more R peaks of the waveform are used to gate a cardiac scan to scan only during pre-determined windows within a cardiac phase of interest (prospective gating) or to locate project data corresponding to that window and phase after a scan in the projection data produced thereby (retrospective gating).

[0049] The invention has been described with reference to the preferred embodiments. Modifications and alterations may occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be constructed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

1. A static charge filter for removing static charge in a cardiac electrical signal, comprising:

- a first amplifier configured to amplify an input signal, which includes the cardiac electrical signal and static charge from an electrode, which is in a path of an X-ray beam;
 - a limiter configured to limit a maximum voltage of the signal based on a predetermined clamping threshold, producing a voltage clamped signal;
 - a filter configured to filter high frequency components of the voltage clamped signal, producing a filtered signal; and
 - a second amplifier configured to scale an amplitude of the filtered signal so that cardiac electrical signal in an output signal has a same voltage level as a voltage level of the cardiac electrical signal in the input signal.
2. The static charge filter of claim 1, further comprising: a separate processing channel for each of a plurality of electrodes.
3. The static charge filter of claim 2, wherein only one of the separate processing channels is configured to remove static charge and the other processing channels are pass-throughs.
4. The static charge filter of claim 2, wherein all of the separate processing channels are configured to remove static charge.
5. The static charge filter of claim 2, further comprising: a feedback loop of the first amplifier configured to automatically adjust the first gain so that the signal is at about the predetermined clamping threshold.
6. The static charge filter of claim 1, wherein the limiter comprises:
- an electrical path through the limiter;
 - an electrical ground; and
 - a pair of diodes electrically connected between the path and electrical ground in parallel and in reverse polarity, wherein a spike of the static charge in the signal exceeding the predetermined clamping threshold turns one of the diodes on, clamping a maximum voltage of the signal to the predetermined clamping threshold, reducing the spike.
7. The static charge filter of claim 1, wherein the filter is a clock programmable low pass filter configured to reject the high frequency components out of band harmonics produced by clamping the signal.
8. The static charge filter of claim 1, wherein the filter is a switched capacitor low pass filter.
9. The static charge filter of claim 1, wherein the first gain is a first value A and the second gain is at least a second value 1/A.
10. A system, the comprising:
- a gantry with an examination region;
 - an X-ray radiation source configured to produce and transmit X-ray radiation that traverses the examination region;
 - an X-ray sensitive detector array configured to detect X-ray radiation traversing the examination region and generate projection data;
 - a reconstructor configured to reconstruct at least a portion of the projection data and generate an image indicative thereof; and
 - a static charge filter configured to receive cardiac electrical signals sensed by electrodes in the examination region, wherein at least one of the electrodes is in a path

- of the X-ray radiation traversing the examination region, and remove static charge from the cardiac electrical signal.
- 11.** The system of claim **10**, further comprising: an electrocardiogram monitor configured to process signals output by the static charge filter and produce an electrocardiogram waveform.
- 12.** The system of claim **11**, further comprising: an R peak identifier configured to identify R peaks in the waveform, wherein the identified R peaks control an operation of the system during at least one of a prospectively gated cardiac scan or a retrospectively gated cardiac scan.
- 13.** The system of claim **10**, wherein the static charge filter comprises:
- a processing channel for at least one of the cardiac electrical signals.
- 14.** The system of claim **13**, wherein the processing channel comprises:
- a first amplifier configured to amplify at least one of the cardiac electrical signals by a first gain, producing an amplified signal;
 - a limiter configured to limit a maximum voltage of the amplified signal based on a predetermined clamping threshold, producing a voltage clamped signal;
 - a filter configured to filter high frequency components of the voltage clamped signal, producing a filtered signal; and
 - a second amplifier configured to scale the filtered signal by a second gain so that cardiac electrical signals in an output signal has a same voltage level as a voltage level of the cardiac electrical signal in the input cardiac electrical signals.
- 15.** The system of claim **14**, further comprising: a feedback loop of the first amplifier is configured to automatically adjust the first gain so that the cardiac electrical signal in the amplified signal is at about the predetermined clamping threshold.
- 16.** The system of claim **14**, wherein the limiter comprises:
- an electrical path through the limiter;
 - an electrical ground; and
 - a pair of diodes electrically connected between the path and electrical ground in parallel and in reverse polarity, wherein a spike in the cardiac electrical signal exceeding the predetermined clamping threshold turns one of the diodes on, clamping a maximum voltage of the cardiac electrical signal to the predetermined clamping threshold, reducing the spike in the cardiac electrical signal.
- 17.** A method, comprising:
- receiving a cardiac electrical signal from an electrode in a path of an X-ray beam during an imaging examination, wherein the cardiac electrical signal includes static charge;
 - removing a static charge voltage spike from the received cardiac electrical signal, producing a static charge filtered cardiac electrical signal;
 - transmitting the static charge filtered cardiac electrical signal to an EKG monitor, which processes the static charge filtered cardiac electrical signal and produces an EKG waveform representing an electrical activity of a heart of the subject.
- 18.** The method of claim **17**, wherein removing the voltage spike comprises:
- amplifying the received cardiac electrical signal by a first gain, producing an amplified signal;
 - clamping a maximum voltage of the amplified signal based on a predetermined clamping threshold, producing a voltage clamped signal;
 - filtering high frequency out of band harmonics of the clamped signal, producing a filtered signal; and
 - scaling an amplitude of the filtered signal using a second gain, which produces an output signal having a voltage level at a voltage level of the input cardiac electrical signal.
- 19.** The method of claim **17**, further comprising:
- identifying an R peak in the EKG waveform; and
 - triggering a scan during an acquisition window of a quiet phase of the heart based on the identified an R peak; and
 - reconstructing projection data from the scan to generate an image of the heart for the acquisition window for the quiet phase.
- 20.** The method of claim **17**, further comprising:
- identifying an R peak in the EKG waveform;
 - identifying a sub-set projection data corresponding to an acquisition window of a quiet phase of the heart from a set of projection data covering more than the acquisition window; and
 - reconstructing the sub-set of projection data to generate an image of the heart for the acquisition window for the quiet phase.

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专利名称(译)	用于静电荷的心脏电信号的静电荷滤波器被路由到心电图仪 (EKG) 监视器		
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摘要(译)

静电荷滤波器 (522) 去除心脏电信号中的静电荷。静电荷滤波器包括第一放大器 (608) , 其被配置为放大输入信号, 该输入信号包括来自电极的心脏电信号和静电荷, 该电极位于X射线束的路径中。静电荷滤波器还包括限制器 (614) , 其被配置为基于预定的钳位阈值限制信号的最大电压, 从而产生电压钳位信号。静电荷滤波器还包括滤波器 (616) , 其被配置为对电压钳位信号的高频分量进行滤波, 从而产生滤波信号。静电荷滤波器还包括第二放大器 (620) , 其被配置为缩放滤波信号的幅度, 使得输出信号中的心脏电信号具有与输入信号中的心脏电信号的电压电平相同的电压电平。

