(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
28 July 2016 (28.07.2016)

(10) International Publication Number
WO 2016/118960 A1

(51) International Patent Classification:
G01T 1/36 (2006.01) *G01T 1/15* (2006.01)
G01T 1/29 (2006.01)

(21) International Application Number:
PCT/US2016/014769

(22) International Filing Date:
25 January 2016 (25.01.2016)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
62/107,036 23 January 2015 (23.01.2015) US
62/150,887 22 April 2015 (22.04.2015) US

(71) Applicant: **RENSSELAER POLYTECHNIC INSTITUTE** [US/US]; 110 8th Street, J Building, Troy, NY 12180 (US).

(72) Inventors: **WANG, Ge**; 20 Hilander Dr., Loudonville, NY 12211 (US). **CONG, Wenxiang**; 9 Keystone Ct., Albany, NY 12201 (US). **SHI, Zaifeng**; 29 Latham Village Lane, Apt. 12, Latham, NY 12110 (US).

(74) Agents: **FRANK, Louis, C.** et al.; Saliwanchik Lloyd & Eisenschenk, PO Box 142950, Gainesville, FL 32614-2950 (US).

(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IR, IS, JP, KE, KG, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, ST, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:

— *with international search report (Art. 21(3))*

(54) Title: SPECTRAL X-RAY DETECTORS WITH DYNAMIC ELECTRONIC CONTROL AND COMPUTATIONAL METHODS

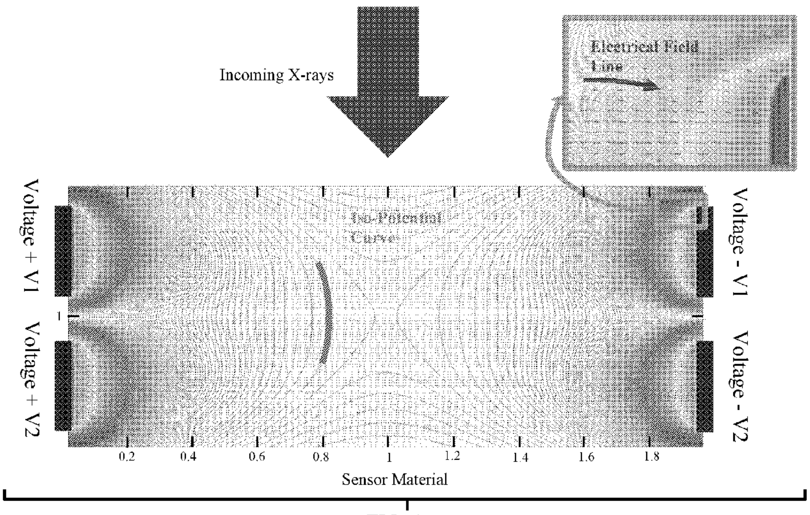


FIG. 1A

(57) **Abstract**: Novel and advantageous methods and systems for performing computed tomography (CT) imaging are disclosed. Electrodes can be connected to appropriate surface sites of a detector element of a CT scanner to capture nearby electron-hole pairs generated by X-rays received on the detector element. This detection can be performed in current- integrating/energy -integrating mode.

WO 2016/118960 A1

DESCRIPTION

SPECTRAL X-RAY DETECTORS WITH DYNAMIC ELECTRONIC CONTROL AND COMPUTATIONAL METHODS

CROSS-REFERENCE TO RELATED APPLICATION

This application claims the benefit of U.S. Provisional Application Serial No. 62/107,036, filed January 23, 2015, and U.S. Provisional Application Serial No. 62/150,887 filed April 22, 2015, both of which are incorporated herein by reference in their entirety, including any figures, tables, and drawings.

BACKGROUND OF INVENTION

Computed Tomography (CT) is a major tool in diagnostic imaging. X-ray detection technology typically uses energy-integrating detectors that add electrical signals, from interactions between an X-ray beam and a material of the detector, over the whole spectrum. Energy-integrating detectors often lose spectral information. Spectral CT (SCT) has advantages over conventional CT by offering detailed spectral information for material decomposition. SCT can also reduce beam-hardening artifacts and radiation dose. However, related art SCT is slower, less stable, and much more expensive than conventional CT.

Conventional CT is based on energy-integrating and/or current-integrating detectors for data acquisition. With photon-counting detectors, SCT can offer additional spectral information for diagnosis, such as discriminating tissues and differentiating calcium and iodine. Related art commercial dual-energy CT technologies include a dual-source CT system from Siemens, a dual-kVp system from GE, and a dual-layer-detector-based system from Philips. These scanners are not for SCT systems because only two material basis functions can be extracted.

BRIEF SUMMARY

The subject invention provides novel and advantageous methods and systems for performing imaging, such as computed tomography (CT) imaging (e.g., spectral CT (SCT) imaging or energy/current integrating CT imaging). Electrical connectors (e.g., electrodes) can be connected to appropriate surface sites of a detector element of a CT scanner to capture nearby electron-hole pairs generated by X-rays received on the detector element. The spectral information of interacting X-ray photons can be related to the site/depth of the interaction between X-rays and the detector material. This process can be accurately modeled, quantified, and/or inverted, according to the penetration/interaction location of the X-rays inside the detector element (e.g., a semiconductor sensor) for direct X-ray detection. This can be based on, for example, the radiative transport equation or its approximation. This detection can be done in the current-/energy-integrating mode.

In an embodiment, an imaging system can include: a CT scanner including an X-ray source; and a detector for receiving X-ray radiation from the X-ray source after it passes through a sample to be imaged. The detector can include a first pair of electrodes and a second pair of electrodes disposed thereon and configured to provide a first voltage and a second voltage, respectively, to the detector. The detector can include a first layer and a second layer, and the first and second pairs of electrodes can be disposed on and apply the first voltage and the second voltage to the first and second layers, respectively.

In another embodiment, a method of imaging can include: providing X-ray radiation to a sample to be imaged; collecting the X-ray radiation with a detector; providing a first voltage to the detector using a first pair of electrodes disposed thereon; and providing a second voltage to the detector using a second pair of electrodes disposed thereon. The detector can include a first layer and a second layer, and the first and second pairs of electrodes can be disposed on and apply the first voltage and the second voltage to the first and second layers, respectively.

BRIEF DESCRITPION OF THE DRAWINGS

Figure 1A shows a detector with electrodes disposed parallel to the direction of incoming X-rays, and such that the incoming X-rays are first incident on a side of the detector not having any electrodes disposed thereon. The top electrodes (left and right) as depicted in the figure are an electrode pair, and the bottom two electrodes (left and right) as depicted in the figure are an electrode pair. The electric field lines are based on the voltages of both electrode pairs being equal (V1=V2). The inset shows a zoomed-in view of the electric field lines.

Figure 1B shows the detector of Figure 1A but with a close-up view of the electric field lines.

Figure 1C shows a detector with electrodes disposed parallel to the direction of incoming X-rays, and such that the incoming X-rays are first incident on a side of the detector not having any electrodes disposed thereon. The top electrodes (left and right) as depicted in the figure are an electrode pair, and the bottom two electrodes (left and right) as depicted in the figure are an electrode pair. The electric field lines are based on the voltage of the bottom pair being greater than that of the top pair (V2>V1). The inset shows a zoomed-in view of the electric field lines.

Figure 1D shows the detector of Figure 1C but with a close-up view of the electric field lines.

Figure 1E shows a detector with electrodes disposed perpendicular to the direction of incoming X-rays, and such that the incoming X-rays are first incident on a side of the detector having electrodes disposed thereon. The top electrodes (left and right) as depicted in the figure are an electrode pair, and the bottom two electrodes (left and right) as depicted in the figure are an electrode pair. The electric field lines are based on the voltages of both electrode pairs being equal (V1=V2). The inset shows a zoomed-in view of the electric field lines.

Figure 1F shows the detector of Figure 1E but with a close-up view of the electric field lines.

Figure 1G shows a detector with electrodes disposed perpendicular to the direction of incoming X-rays, and such that the incoming X-rays are first incident on a side of the detector having electrodes disposed thereon. The top electrodes (left and right) as depicted in the figure are an electrode pair, and the bottom two electrodes (left and right) as depicted in the figure are an electrode pair. The electric field lines are based on the voltage of the bottom pair being greater than that of the top pair (V2>V1). The inset shows a zoomed-in view of the electric field lines.

Figure 1H shows the detector of Figure 1G but with a close-up view of the electric field lines.

Figure 2 shows a detector having four layers and with four voltages being applied with electrodes, one voltage applied to each layer. Each current (I1, I2, I3, and I4) is that flowing into the respective circle representing the voltage (V1, V2, V3, and V4). The four layers have thicknesses H1, H2, H3, and H4, respectively. The incoming X-rays first go through the first layer, then the second, and so on.

Figure 3 shows a schematic view of direct detection of X-ray CT for third generation geometry. The detector is shown at the bottom under the human subject being imaged.

Figure 4A shows a graphical representation of a ratio of 2:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 4B shows a graphical representation of a ratio of 3:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 4C shows a graphical representation of a ratio of 4:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 5A shows a true image of gadolinium (Gd) contrast agent at a concentration of 0.3 mol/L.

Figure 5B shows a reconstructed image using a ratio of 1:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 5C shows a reconstructed image using a ratio of 2:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 5D shows a reconstructed image using a ratio of 3:1 between number of integrating bins and number of counting bins for a numerical simulation.

Figure 6 shows a schematic view of an X-ray CT scanner.

Figure 7 shows a view of a scanner with an enlarged view of a dual-layer detector, where one layer detects low-energy X-rays and the other layer detects high-energy X-rays.

Figure 8 shows a schematic view demonstrating reading of X-ray photons from a detector.

Figure 9 shows an image obtained from a dual-energy CT scanner.

Figure 10 shows a schematic view of a conventional charge-coupled device (CCD) camera.

Figure 11 shows a schematic view of a spectral CCD camera that can be used according to an embodiment of the subject invention.

Figure 12 shows a schematic view of a spectral CCD camera that can be used according to an embodiment of the subject invention, demonstrating charge generation and collection.

Figure 13 shows a schematic view of a spectral CCD camera that can be used according to an embodiment of the subject invention, demonstrating energy resolving.

Figure 14 shows a table with values of a related art Medipix3 detector and a detector according to an embodiment of the subject invention.

Figure 15 shows a schematic view of a pixelated detector demonstrating reading of X-ray photons from a detector.

Figure 16A shows a schematic view of direct detection of X-ray CT for fourth generation geometry. The detector is shown around the human subject being imaged.

Figure 16B shows a flow diagram demonstrating direct detection of X-ray CT for fourth generation geometry.

Figure 17 shows a two-layer detector used as part of fourth generation geometry. The inset shows a detection plot for the first and second layers. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 18A shows a detection plot at varying energies for fourth generation geometry. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 18B shows a detection plot at varying energies for fourth generation geometry. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 18C shows a detection plot at varying energies for fourth generation geometry. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 18D shows a detection plot at varying energies for fourth generation geometry. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 19 shows a decomposed image of a Gd contrast agent.

Figure 20 shows a detection plot at varying energies based on the inset equation. The y-axis of the detection plot is photons/second.

Figure 21A shows an enlarged version of the water basis component of the image in Figure 19.

Figure 21B shows an enlarged version of the bone basis component of the image in Figure 19.

Figure 21C shows an enlarged version of the Gd basis component of the image in Figure 19.

Figure 22A shows an image of Gd contrast agent at a concentration of 0.06 mol/L (top portion) and a reconstruction of that image (bottom portion).

Figure 22B shows an image of Gd contrast agent at a concentration of 0.09 mol/L (top portion) and a reconstruction of that image (bottom portion).

Figure 22C shows an image of Gd contrast agent at a concentration of 0.12 mol/L (top portion) and a reconstruction of that image (bottom portion).

Figure 22D shows an image of Gd contrast agent at a concentration of 0.15 mol/L (top portion) and a reconstruction of that image (bottom portion).

Figure 23 shows a plot of the root mean square error (RMSE) versus Gd concentration for components of an image of Gd contrast dye.

Figure 24A shows a schematic view of a radiation source providing radiation through a human subject to a detector.

Figure 24B shows two detection plots at varying energy. The y-axis of each detection plot is photons/second.

Figure 25 shows a detection plot at varying energy after 22 cm of water attenuation.

Figure 26A shows a detection plot for first and second layers of an attenuated detector. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 26B shows a detection plot for first and second layers of an attenuated detector. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 26C shows a detection plot for first and second layers of an attenuated detector. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 26D shows a detection plot for first and second layers of an attenuated detector. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Figure 27A shows a detection plot for first, second, and third layers in a three-layer detector.

Figure 27B shows a detection plot for first, second, and third layers in a three-layer detector.

Figure 27C shows a detection plot for first, second, and third layers in a three-layer detector.

Figure 27D shows a detection plot for first, second, and third layers in a three-layer detector.

DETAILED DESCRIPTION

The subject invention provides novel and advantageous methods and systems for performing imaging, such as computed tomography (CT) imaging (e.g., spectral CT (SCT) imaging or energy/current integrating CT imaging). Electrical connectors (e.g., electrodes) can be connected to appropriate surface sites of a detector element of a CT scanner to capture nearby electron-hole pairs generated by X-rays received on the detector element. The spectral information of interacting X-ray photons can be related to the site/depth of the interaction between X-rays and the detector material. This process can be accurately modeled, quantified, and/or inverted, according to the penetration/interaction location of the X-rays inside the detector element (e.g., a semiconductor sensor) for direct X-ray detection. This can be based on, for example, the radiative transport equation or its approximation. This detection can be done in the current-/energy-integrating mode.

Systems and methods of the subject invention can electronically control the collection of spectral information in the current-integration mode on a detector element (e.g., during CT scanning). This can be used with many detection schemes, including dual-layer detectors, SCT detectors, and hybrid detectors.

The voltage(s) applied to the electrodes connected to the detector element can be changed. By changing the voltage(s) applied to the electrodes (or metal plates), the electric field inside the sensor (detector element) can be modified to define various current-conducting layers. For example, two sets of electrodes can be used, and the voltage applied across each set can be the same or different. The voltages can be modified during detection as well. The set of electrodes across which a higher voltage is applied can detect more charges than the same loop driven by a lower voltage. In this way, different spectral ratios between two layers can be obtained for X-ray spectral sensing. This dynamic variation of electrode voltage is very flexible and cost-effective in practice.

The detector configuration having electrodes connected thereto for voltage modification/modulation is quite different and advantageous compared to related art scanners, such as the dual-layer detector design (e.g., that used in the Philips CT scanner). In embodiments of the subject invention, direct X-ray detection can be used such that data acquisition can be dynamically modulated by voltage into higher and lower bins in a desirable ratio. Any ratio can be used, such as 2:1 (higher energy bin : lower energy bin), 3:1, 4:1, 1:2, 1:3, 1:4, 5:1, 1:5, 1.5:1, 1:1.5. These ratios are for exemplary purposes and should not be construed as limiting. Embodiments of the subject invention can be used in a detector having any number of layers, such as two layers (dual layer), three layers (triple layer), four layers (quadruple layer), five layers, or more. A first layer can be made sensitive to low-energy photons while letting most of the high-energy photons penetrate through to subsequent layers. In certain embodiments, the multiple layers can all be the same material (e.g., the material of the detector element sensor) and can be defined by electrode sets such that a first electrode set defines a first layer, a second electrode set defines a second layer, etc.

The detector can include a sensor material, including at least one of cadmium zinc telluride (CZT), silicon (e.g., p^-, p^+, n^-, or n^+ silicon), and cadmium telluride (CdTe), though embodiments are not limited thereto. In an embodiment, the detector sensor material is silicon. In a further embodiment, the detector sensor material is p^- silicon In another embodiment, the detector sensor material is CZT.

In an embodiment, a dual-layer detector can break the X-ray spectrum into low and high energies at different proportions during detection. For example, the low and high energies can be broken into proportions of 10%/90%, 20%/80%, 70%/30%, 60%/40%, 50%/50%, 40%/60%, 30%,70%, 20%/80%, and 10%/90%. These proportions are for exemplary purposes and should not be construed as limiting. In a further embodiment, these detector variants can be distributed in a natural sequence and repeated until a full detector ring is covered in third generation geometry. The breaking of the X-ray spectrum can be performed by changing the applied voltages and/or by using software to read the detected radiation. This breaking of the X-ray spectrum can be applied to multi-layer detectors with more than two layers as well.

Related-art dual-layer detectors use fixed layers and cannot be adapted well for many applications, and related art photon-counting (SCT) detectors are slow and expensive. Embodiments of the subject invention advantageously fill in a gap between these two types of detectors, making up for disadvantages of both. Systems and methods of the subject invention can provide significantly improved performance compared to related art dual-energy CT scanners (approaching SCT performance) while avoiding the high cost and rate limitations of photon-counting (SCT) detectors. Systems and methods of the subject invention therefore have many uses in medical imaging.

Figures 6 and 7 show schematic views of X-ray CT scanners, with Figure 7 including an enlarged view of a dual-layer detector, where one layer detects low-energy X-rays and the other layer detects high-energy X-rays. Figure 8 demonstrates reading of X-ray photons from a detector, and Figure 9 shows an image obtained from a dual-energy CT scanner. **Figure 15** shows a schematic view of a pixelated detector demonstrating reading of X-ray photons from a detector, and Figure 16A shows a schematic view of direct detection of X-ray CT for fourth generation geometry. The detector is shown around the human subject being imaged. Figure 16B shows a flow diagram demonstrating direct detection of X-ray CT for fourth generation geometry, and Figure 17 shows a two-layer detector used as part of fourth generation geometry. The inset of Figure 17 shows a detection plot for the first and second layers. The y-axis of the detection plot is photons/second. The first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot.

Systems and methods of the subject invention can electronically control multi-layer detectors (e.g., dual layer detectors) to acquire X-ray photons in two (or more) energy bins, which can be effectively changed under voltage control. Figures 1A-1H show detectors according to embodiments of the subject invention. Referring to Figures 1A-1H, a detector can include at least one electrode set for applying a voltage. The electrodes can be disposed in various configurations. In many embodiments, at least one set of electrodes is provided for each layer of the detector. The set of electrodes can define the layers of the detector. In a further embodiment, at least two sets of electrodes can be provided for each layer of the detector.

Figure 1A shows a detector including electrodes disposed parallel to the direction of incoming X-rays, in positions such that the incoming X-rays are first incident on a side of the detector not having any electrodes disposed thereon. The top electrodes (left and right) as depicted in Figure 1A are an electrode pair, and the bottom two electrodes (left and right) as depicted in Figure 1A are an electrode pair. The electric field lines are based on the voltages of both electrode pairs being equal (V1=V2). The inset shows a zoomed-in view of the electric field lines. Figure 1B shows the detector of Figure 1A but with a close-up view of the electric field lines.

Figure 1C shows a detector with electrodes disposed parallel to the direction of incoming X-rays, in positions such that the incoming X-rays are first incident on a side of the detector not having any electrodes disposed thereon. The top electrodes (left and right) as depicted in Figure 1C are an electrode pair, and the bottom two electrodes (left and right) as depicted in Figure 1C are an electrode pair. The electric field lines are based on the voltage of the bottom pair being greater than that of the top pair (V2>V1). The inset shows a zoomed-in view of the electric field lines. Figure 1D shows the detector of Figure 1C but with a close-up view of the electric field lines.

Figure 1E shows a detector with electrodes disposed perpendicular to the direction of incoming X-rays, in positions such that the incoming X-rays are first incident on a side of the detector having electrodes disposed thereon. The top electrodes (left and right) as depicted in Figure 1E are an electrode pair, and the bottom two electrodes (left and right) as depicted in Figure 1E are an electrode pair. The electric field lines are based on the voltages of both electrode pairs being equal (V1=V2). The inset shows a zoomed-in view of the electric field lines. Figure 1F shows the detector of Figure 1E but with a close-up view of the electric field lines.

Figure 1G shows a detector with electrodes disposed perpendicular to the direction of incoming X-rays, in positions such that the incoming X-rays are first incident on a side of the detector having electrodes disposed thereon. The top electrodes (left and right) as depicted in Figure 1G are an electrode pair, and the bottom two electrodes (left and right) as depicted in Figure 1G are an electrode pair. The electric field lines are based on the voltage of the bottom pair being greater than that of the top pair (V2>V1). The inset shows a zoomed-in view of the electric field lines. Figure 1H shows the detector of Figure 1G but with a close-up view of the electric field lines.

Figure 2 shows a detector having four layers and with four voltages being applied with electrodes, one voltage applied to each layer. Each current (I1, I2, I3, and I4) is that flowing into the respective circle representing the voltage (V1, V2, V3, and V4). The four layers have thicknesses H1, H2, H3, and H4, respectively. The incoming X-rays first go through the first layer (having thickness H1), then the second, and so on. The thicknesses

(H1, H2, H3, and H4) can be all the same, all different, or some can be the same as at least one other while others are different from at least one other.

Referring to Figure 2, depending on the relative strengths of paired driving voltages V1, V2, V3, and V4, the loop currents I1, I2, I3, and I4 can be solved using Kirchhoff's voltage law (KVL) or Kirchhoff's current law (KCL), both of which are well-known within the art. These data can be used to compute interaction rates in each of the four layers of thicknesses H1, H2, H3, and H4. These interaction rates can be converted to data inside four energy windows in the current-integrating mode of the detector. This allows electrodes to be placed within the detector material and can be extended to other configurations as well.

In embodiments of the subject invention, electrical connectors (e.g., electrode sets or electrode pairs) can be connected to appropriate surface sites of the detector to capture nearby electron-hole pairs. The energy information can be related to the site/depth of the interaction between X-rays and the detector material. In certain embodiments, this process can be modeled, quantified, and/or inverted, according to the penetration depth of the X-ray(s) into the detector material (e.g., semiconductor sensor material) for direct X-ray detection. This can be based, for example, on the radiative transport equation or its approximation.

The voltages applied to the metallic plates/electrodes can be changed, thereby modifying the electric field to define various current-conducting layers. For example, the electrodes across which a higher voltage is applied can detect more charges than those across which a lower is applied (i.e., a lower voltage loop). This can lead to the obtaining of different spectral ratios (as disclosed herein) for X-ray spectral sensing. This dynamic variation of electrode voltage is very flexible and cost-effective in practice.

X-ray detection technology includes energy-integration detection and photon-counting detection. Related art X-ray scanners typically use energy-integrating detectors where electrical signals, from interactions between an X-ray beam and materials, are added up over the whole spectrum. In contrast, photon-counting detectors count photons with energy bins. Photon-counting detectors have advantages relative to energy-integrating detectors are also slower and more expensive. The systems and methods described herein can be considered different from energy-integrating detectors and photon-counting detectors by having the ability to record spectral CT data with voltage-controlled layers. These systems and methods have many practical applications, including but not limited to preclinical imaging, clinical imaging, security screening, and industrial evaluation.

This application shares some aspects with International Patent Application No. PCT/US2015/067441, filed December 22, 2015, and U.S. Provisional Application Serial No. 62/095,235, filed December 22, 2014, both of which are hereby incorporated herein by reference in their entirety, including any figures, tables, and drawings (see also Figures 10-13 of the subject application).

The methods and processes described herein can be embodied as code and/or data. The software code and data described herein can be stored on one or more computer-readable media, which may include any device or medium that can store code and/or data for use by a computer system. When a computer system reads and executes the code and/or data stored on a computer-readable medium, the computer system performs the methods and processes embodied as data structures and code stored within the computer-readable storage medium.

It should be appreciated by those skilled in the art that computer-readable media include removable and non-removable structures/devices that can be used for storage of information, such as computer-readable instructions, data structures, program modules, and other data used by a computing system/environment. A computer-readable medium includes, but is not limited to, volatile memory such as random access memories (RAM, DRAM, SRAM); and non-volatile memory such as flash memory, various read-only-memories (ROM, PROM, EPROM, EEPROM), magnetic and ferromagnetic/ferroelectric memories (MRAM, FeRAM), and magnetic and optical storage devices (hard drives, magnetic tape, CDs, DVDs); network devices; or other media now known or later developed that is capable of storing computer-readable information/data. Computer-readable media should not be construed or interpreted to include any propagating signals. A computer-readable medium of the subject invention can be, for example, a compact disc (CD), digital video disc (DVD), flash memory device, volatile memory, or a hard disk drive (HDD), such as an external HDD or the HDD of a computing device, though embodiments are not limited thereto. A computing device can be, for example, a laptop computer, desktop computer, server, cell phone, or tablet, though embodiments are not limited thereto.

When the term “about” is used herein, in conjunction with a numerical value, it is understood that the value can be in a range of 95% of the value to 105% of the value, i.e. the value can be +/- 5% of the stated value. For example, “about 1 kg” means from 0.95 kg to 1.05 kg.

The subject invention includes, but is not limited to, the following exemplified embodiments.

Embodiment 1. An imaging system, comprising:

a computed tomography (CT) scanner including an X-ray source; and

a detector for receiving X-ray radiation from the X-ray source after it passes through a sample to be imaged,

wherein the detector includes a first pair of electrodes and a second pair of electrodes disposed thereon and configured to provide a first voltage and a second voltage, respectively, to the detector.

Embodiment 2. The imaging system according to embodiment 1, wherein the detector includes a first layer and a second layer.

Embodiment 3. The imaging system according to embodiment 2, wherein the first pair of electrodes is disposed on the first layer and applies the first voltage to the first layer, and

wherein the second pair of electrodes is disposed on the second layer and applies the second voltage to the second layer.

Embodiment 4. The imaging system according to any of embodiments 2-3, wherein the first layer and the second layer comprise the same material.

Embodiment 5. The imaging system according to any of embodiments 1-4, wherein the detector comprises a sensor material.

Embodiment 6. The imaging system according to any of embodiments 1-5, wherein the detector comprises at least one of cadmium zinc telluride (CZT), silicon (e.g., p^-, p^+, n^-, or n^+ silicon), and cadmium telluride (CdTe).

Embodiment 7. The imaging system according to any of embodiments 1-6, wherein the detector comprises CZT.

Embodiment 8. The imaging system according to any of embodiments 1-6, wherein the detector comprises CdTe.

Embodiment 9. The imaging system according to any of embodiments 1-6, wherein the detector comprises silicon.

Embodiment 10. The imaging system according to any of embodiments 1-6, wherein the detector comprises p^-, p^+, n^-, or n^+ silicon.

Embodiment 11. The imaging system according to any of embodiments 1-6, wherein the detector comprises p^- silicon.

Embodiment 12. The imaging system according to any of embodiments 1-6, wherein the detector comprises p^+ silicon.

Embodiment 13. The imaging system according to any of embodiments 1-6, wherein the detector comprises n^- silicon.

Embodiment 14. The imaging system according to any of embodiments 1-6, wherein the detector comprises n^+ silicon.

Embodiment 15. The imaging system according to any of embodiments 1-14, wherein the first voltage is equal to the second voltage.

Embodiment 16. The imaging system according to any of embodiments 1-14, wherein the first voltage is greater than the second voltage.

Embodiment 17. The imaging system according to any of embodiments 1-14, wherein the first voltage is less than the second voltage.

Embodiment 18. The imaging system according to any of embodiments 2-17, wherein the first layer and the second layer are defined by the first and second pairs of electrodes, respectively.

Embodiment 19. The imaging system according to any of embodiments 1-18, wherein both electrodes of the first pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

Embodiment 20. The imaging system according to any of embodiments 1-19, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector not having any of the first pair of electrodes disposed thereon.

Embodiment 21. The imaging system according to any of embodiments 1-20, wherein both electrodes of the second pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

Embodiment 22. The imaging system according to any of embodiments 1-21, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector not having any of the second pair of electrodes disposed thereon.

Embodiment 23. The imaging system according to any of embodiments 1-18 and 20-22, wherein both electrodes of the first pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

Embodiment 24. The imaging system according to any of embodiments 1-19 and 21-23, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector having both electrodes of the first pair of electrodes disposed thereon.

Embodiment 25. The imaging system according to any of embodiments 1-20 and 22-24, wherein both electrodes of the second pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

Embodiment 26. The imaging system according to any of embodiments 1-21 and 23-25, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector having both electrodes of the second pair of electrodes disposed thereon.

Embodiment 27. The imaging system according to any of embodiments 1-26, wherein the detector further comprises a third pair of electrodes disposed thereon and configured to provide a third voltage to the detector.

Embodiment 28. The imaging system according to embodiment 27, wherein the detector further comprises a fourth pair of electrodes disposed thereon and configured to provide a fourth voltage to the detector.

Embodiment 29. The imaging system according to any of embodiments 2-28, wherein the detector further comprises a third layer.

Embodiment 30. The imaging system according to embodiment 29, wherein the detector further comprises a fourth layer.

Embodiment 31. The imaging system according to embodiment 27, wherein the detector includes first to third layers, and wherein the first to third pairs of electrodes are disposed on and configured to provide the first to third voltages to the first to third layers, respectively.

Embodiment 32. The imaging system according to embodiment 28, wherein the detector includes first to fourth layers, and wherein the first to fourth pairs of electrodes are disposed on and configured to provide the first to fourth voltages to the first to fourth layers, respectively.

Embodiment 33. The imaging system according to embodiment 28, wherein the first to fourth voltages are all the same.

Embodiment 34. The imaging system according to embodiment 28, wherein the first to fourth voltages are all different from each other.

Embodiment 35. The imaging system according to any of embodiments 30 and 32, wherein the detector comprises the third and fourth pairs of electrodes, and wherein the first, second, third, and fourth layers are defined by the first to fourth pairs of electrodes, respectively.

Embodiment 36. The imaging system according to any of embodiments 1-35, further comprising a controller configured to control the first and second pairs of electrodes (and third and fourth pairs of electrodes, if present) to apply the first and second voltages (and the third and fourth voltages, if the third and fourth pairs of electrodes are present) such that photons in the detector are captured in a high energy bin and a low energy bin.

Embodiment 37. The imaging system according to embodiment 36, wherein the controller is configured to control the pairs of electrodes to apply the voltages such that photons in the detector are captured in the high energy bin and the low energy bin in a predetermined ratio.

Embodiment 38. The imaging system according to any of embodiments 2-37, wherein a thickness of each layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is the same as that of each other layer of the detector present.

Embodiment 39. The imaging system according to any of embodiments 2-37, wherein a thickness of at least one layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is different from that of at least one other layer of the detector present.

Embodiment 40. The imaging system according to any of embodiments 2-37, wherein a thickness of each layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is the different from that of each other layer of the detector present.

Embodiment 41. The imaging system according to any of embodiments 1-40, further comprising a (non-transitory) machine-readable medium (e.g., a computer-readable medium) having machine-executable (e.g., computer-executable) instructions for performing an energy resolving process on the collected X-ray radiation (e.g., on collected charges of the X-ray radiation).

Embodiment 42. The imaging system according to embodiment 41, wherein the energy resolving process includes:

determining the generated charge density within the detector using Formula 1:

$$\text{Formula 1 - } m_1E_1N_1a_{k1} + m_2E_2N_2a_{k2} + \cdots + m_nE_nN_na_{kn} = g_k(x),$$

where E_i is photon energy, N_i is photon density with energy E_i, a_{ki} is the attenuation coefficient of photon with energy E_i for the given material thickness, m_i is an empirical coefficient that represents the number of generated charges by photons with energy E_i per energy unit, and $g_k(x)$ is the generated charge density within the material of the detector of a specific thickness; and

repeating the determination of the generated charge density at a different thickness within the material of the detector.

Embodiment 43. The imaging system according to any of embodiments 41-42, further comprising a processer, wherein the energy resolving process is performed by the processor.

Embodiment 44. The imaging system according to any of embodiments 1-43, configured to perform imaging on a sample that is a part of a human patient (e.g., a body part).

Embodiment 45. The imaging system according to any of embodiments 1-44, wherein the X-ray source is configured to provide X-ray radiation that has an energy of from 10 keV to 120 keV.

Embodiment 46. The imaging system according to any of embodiments 1-44, wherein the X-ray source is configured to provide X-ray radiation that has an energy of less than 20 keV.

Embodiment 47. The imaging system according to any of embodiments 1-44, wherein the X-ray source is configured to provide X-ray radiation that has an energy of more than 20 keV.

Embodiment 48. The imaging system according to any of embodiments 1-47, wherein the detector includes a fixed thresholding detector.

Embodiment 49. The imaging system according to any of embodiments 1-47, wherein the detector includes a dynamic thresholding detector.

Embodiment 50. The imaging system according to any of embodiments 1-49, wherein the CT scanner (potentially in combination with the detector) has third-generation geometry.

Embodiment 51. The imaging system according to any of embodiments 1-49, wherein the CT scanner (potentially in combination with the detector) has fourth-generation geometry.

Embodiment 52. The imaging system according to any of embodiments 1-51, further comprising an analog-digital converter (ADC) electrically connected to the detector and configured to quantify collected charges from the detector.

Embodiment 53. A method of imaging, comprising:

providing the imaging system according to any of embodiments 1-52; and

using the imaging system for its intended purpose to image the sample.

Embodiment 54. A method of imaging, comprising:

providing X-ray radiation to a sample to be imaged;

collecting the X-ray radiation with a detector;

providing a first voltage to the detector using a first pair of electrodes disposed thereon; and

providing a second voltage to the detector using a second pair of electrodes disposed thereon.

Embodiment 55. The method according to embodiment 54, wherein the detector includes a first layer and a second layer.

Embodiment 56. The method according to embodiment 55, wherein the first pair of electrodes is disposed on the first layer and applies the first voltage to the first layer, and

wherein the second pair of electrodes is disposed on the second layer and applies the second voltage to the second layer.

Embodiment 57. The method according to any of embodiments 55-56, wherein the first layer and the second layer comprise the same material.

Embodiment 58. The method according to any of embodiments 54-57, wherein the detector comprises a sensor material.

Embodiment 59. The method according to any of embodiments 54-58, wherein the detector comprises at least one of cadmium zinc telluride (CZT), silicon (e.g., p^-, p^+, n^-, or n^+ silicon), and cadmium telluride (CdTe).

Embodiment 60. The method according to any of embodiments 54-59, wherein the detector comprises CZT.

Embodiment 61. The method according to any of embodiments 54-59, wherein the detector comprises CdTe.

Embodiment 62. The method according to any of embodiments 54-59, wherein the detector comprises silicon.

Embodiment 63. The method according to any of embodiments 54-59, wherein the detector comprises p^-, p^+, n^-, or n^+ silicon.

Embodiment 64. The method according to any of embodiments 54-59, wherein the detector comprises p^- silicon.

Embodiment 65. The method according to any of embodiments 54-59, wherein the detector comprises p^+ silicon.

Embodiment 66. The method according to any of embodiments 54-59, wherein the detector comprises n^- silicon.

Embodiment 67. The method according to any of embodiments 54-59, wherein the detector comprises n^+ silicon.

Embodiment 68. The method according to any of embodiments 54-67, wherein the first voltage is equal to the second voltage.

Embodiment 69. The method according to any of embodiments 54-67, wherein the first voltage is greater than the second voltage.

Embodiment 70. The method according to any of embodiments 54-67, wherein the first voltage is less than the second voltage.

Embodiment 71. The method according to any of embodiments 54-70, further comprising dynamically modulating at least one of the first voltage and the second voltage such that photons in the detector are captured in a high energy bin and a low energy bin in a predetermined ratio.

Embodiment 72. The method according to any of embodiments 54-70, further comprising dynamically modulating both the first voltage and the second voltage (and any other voltages that may be applied by additional electrode pairs that may be present) such that photons in the detector are captured in a high energy bin and a low energy bin in a predetermined ratio.

Embodiment 73. The method according to any of embodiments 71-72, wherein the ratio (higher energy bin : lower energy bin) is 2:1, 3:1, 4:1, 1:2, 1:3, 1:4, 5:1, 1:5, 1.5:1, 1:1.5, 2.5:1, 1:2.5, 3.5:1, 1:3.5, 4.5:1, or 1:4.5.

Embodiment 74. The method according to any of embodiments 54-73, further comprising dividing the X-ray spectrum into low and high energies at different proportions during detection such that detected X-ray radiation is classified as either high energy or low energy during detection by modulating the first and second voltages (and any other voltages that may be applied by additional electrode pairs that may be present).

Embodiment 75. The method according to embodiment 74, wherein the low and high energies are divided into proportions of (low/high) 10%/90%, 20%/80%, 70%/30%, 60%/40%, 50%/50%, 40%/60%, 30%,70%, 20%/80%, and 10%/90% during detection by modulating the first and second voltages (and any other voltages that may be applied by additional electrode pairs that may be present).

Embodiment 76. The method according to any of embodiments 74-75, further comprising distributing these detector variants in a natural sequence, and repeating until a full detector ring is covered in third generation geometry.

Embodiment 77. The method according to any of embodiments 74-75, further comprising distributing these detector variants in a natural sequence, and repeating until a full detector ring is covered in fourth generation geometry.

Embodiment 78. The method according to any of embodiments 55-77, wherein the first layer and the second layer are defined by the first and second pairs of electrodes, respectively.

Embodiment 79. The method according to any of embodiments 54-78, wherein both electrodes of the first pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

Embodiment 80. The method according to any of embodiments 54-79, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector not having any of the first pair of electrodes disposed thereon.

Embodiment 81. The method according to any of embodiments 54-80, wherein both electrodes of the second pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

Embodiment 82. The method according to any of embodiments 54-81, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector not having any of the second pair of electrodes disposed thereon.

Embodiment 83. The method according to any of embodiments 54-78 and 79-82, wherein both electrodes of the first pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

Embodiment 84. The method according to any of embodiments 54-79 and 81-83, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector having both electrodes of the first pair of electrodes disposed thereon.

Embodiment 85. The method according to any of embodiments 54-80 and 82-84, wherein both electrodes of the second pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

Embodiment 86. The method according to any of embodiments 54-81 and 83-85, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector having both electrodes of the second pair of electrodes disposed thereon.

Embodiment 87. The method according to any of embodiments 54-86, wherein the detector further comprises a third pair of electrodes disposed thereon and configured to provide a third voltage to the detector.

Embodiment 88. The method according to embodiment 87, wherein the detector further comprises a fourth pair of electrodes disposed thereon and configured to provide a fourth voltage to the detector.

Embodiment 89. The method according to any of embodiments 55-88, wherein the detector further comprises a third layer.

Embodiment 90. The method according to embodiment 89, wherein the detector further comprises a fourth layer.

Embodiment 91. The method according to embodiment 87, wherein the detector includes first to third layers, and wherein the first to third pairs of electrodes are disposed on and configured to provide the first to third voltages to the first to third layers, respectively.

Embodiment 92. The method according to embodiment 88, wherein the detector includes first to fourth layers, and wherein the first to fourth pairs of electrodes are disposed on and configured to provide the first to fourth voltages to the first to fourth layers, respectively.

Embodiment 93. The method according to embodiment 88, wherein the first to fourth voltages are all the same.

Embodiment 94. The method according to embodiment 88, wherein the first to fourth voltages are all different from each other.

Embodiment 95. The method according to any of embodiments 90 and 92, wherein the detector comprises the third and fourth pairs of electrodes, and wherein the first, second, third, and fourth layers are defined by the first to fourth pairs of electrodes, respectively.

Embodiment 96. The method according to any of embodiments 54-95, further comprising controlling (e.g., using a controller) the first and second pairs of electrodes (and third and fourth pairs of electrodes, if present) to apply the first and second voltages (and the third and fourth voltages, if the third and fourth pairs of electrodes are present) such that photons in the detector are captured in a high energy bin and a low energy bin.

Embodiment 97. The method according to embodiment 96, wherein the pairs of electrodes are controlled to apply the voltages such that photons in the detector are captured in the high energy bin and the low energy bin in a predetermined ratio.

Embodiment 98. The method according to any of embodiments 55-97, wherein a thickness of each layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is the same as that of each other layer of the detector present.

Embodiment 99. The method according to any of embodiments 55-97, wherein a thickness of at least one layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is different from that of at least one other layer of the detector present.

Embodiment 100. The method according to any of embodiments 55-97, wherein a thickness of each layer of the detector present (e.g., first layer, second layer, third layer (if present), fourth layer (if present)) is the different from that of each other layer of the detector present.

Embodiment 101. The method according to any of embodiments 54-100, further comprising performing an energy resolving process on the collected X-ray radiation (e.g., on collected charges of the X-ray radiation).

Embodiment 102. The method according to embodiment 101, wherein the energy resolving process includes:

determining the generated charge density within the detector using Formula 1:

$$\text{Formula 1 - } m_1E_1N_1a_{k1} + m_2E_2N_2a_{k2} + \dots + m_nE_nN_na_{kn} = g_k(x),$$

where E_i is photon energy, N_i is photon density with energy E_i, a_{ki} is the attenuation coefficient of photon with energy E_i for the given material thickness, m_i is an empirical coefficient that represents the number of generated charges by photons with energy E_i per energy unit, and $g_k(x)$ is the generated charge density within the material of the detector of a specific thickness; and

repeating the determination of the generated charge density at a different thickness within the material of the detector.

Embodiment 103. The method according to any of embodiments 101-102, wherein the energy resolving process is performed by a processor.

Embodiment 104. The method according to any of embodiments 54-103, wherein the sample to be imaged is a part of a human patient (e.g., a body part).

Embodiment 105. The method according to any of embodiments 54-104, wherein the X-ray radiation has an energy of from 10 keV to 120 keV.

Embodiment 106. The method according to any of embodiments 54-104, wherein the X-ray radiation has an energy of less than 20 keV.

Embodiment 107. The method according to any of embodiments 54-104, wherein the X-ray radiation has an energy of more than 20 keV.

Embodiment 108. The method according to any of embodiments 54-107, wherein the detector includes a fixed thresholding detector.

Embodiment 109. The method according to any of embodiments 54-107, wherein the detector includes a dynamic thresholding detector.

Embodiment 110. The method according to any of embodiments 101-109, wherein the steps of the energy resolving process are stored on a (non-transitory) machine-readable medium (e.g., a computer-readable medium).

Embodiment 111. The method according to any of embodiments 54-110, wherein the imaging is a computed tomography (CT) scan.

Embodiment 112. The method according to any of embodiments 54-111, wherein the X-ray radiation is provided by an X-ray source of a CT scanner.

Embodiment 113. The method according to embodiment 112, wherein the CT scanner (potentially in combination with the detector) has third-generation geometry.

Embodiment 114. The method according to embodiment 112, wherein the CT scanner (potentially in combination with the detector) has fourth-generation geometry.

Embodiment 115. The method according to any of embodiments 54-114, wherein the detector is placed in an edge-on fashion during imaging, such that the X-ray irradiation enters a side of a substrate of the detector.

Embodiment 116. The imaging system according to any of embodiments 1-52 or the method according to any of embodiments 53-115, wherein each electrode present is a metal electrode.

A greater understanding of the present invention and of its many advantages may be had from the following examples, given by way of illustration. The following examples are illustrative of some of the methods, applications, embodiments and variants of the present invention. They are, of course, not to be considered as limiting the invention. Numerous changes and modifications can be made with respect to the invention.

EXAMPLE 1

A numerical simulation was performed using an algorithm as described by De Man et al. (An Iterative Maximum-Likelihood Polychromatic Algorithm for CT, IEEE Transactions on Medical Imaging, Vol. 20, No. 10, October 2001, which is hereby incorporated herein by reference in its entirety). The simulation was performed on a detector utilizing voltage-controlled layers as described herein. The source-to-center distance was 50 cm, the detector-to-center distance was 50 cm, the detector array width was 60 cm, the sample size was 38 by 38 cm^2, and the a hybrid ratio between number of integrating bins and number of counting bins was utilized, and Figures 4A, 4B, and 4C show graphical representations of ratios of 3:1, 4:1, and 5:1, respectively, between number of integrating bins and number of counting bins. The detection geometry was that as shown in Figure 3.

Figure 5A shows a true image of the gadolinium (Gd) contrast agent at a concentration of 0.3 mol/L used for the simulation. Figure 5B shows a reconstructed image using a ratio of 1:1 between number of integrating bins and number of counting bins, Figure 5C shows a reconstructed image using a ratio of 2:1 between number of integrating bins and number of counting bins, and Figure 5D shows a reconstructed image using a ratio of 3:1 between number of integrating bins and number of counting bins.

EXAMPLE 2

A detector utilizing voltage-controlled layers as described herein was constructed, and a numerical simulation was performed to compare the detector to the Medipix 3 related art detector. The results are shown in Figure 14. Referring to Figure 14, the detector of the subject invention has many advantages, including shorter acquisition time, greater frame rate, and multiple working modes.

EXAMPLE 3

Different configurations of dual-layer detectors according to embodiments of the subject invention were tested for performance of photon counting at varying energies of the X-ray radiation.

Figure 18A shows a detection plot for a control with channel 4n+0, a first layer of 20 mm silicon, a second layer of 20 mm silicon, and no electrodes. Figure 18B shows a detection plot for a first variant with channel 4n+1, a first layer of 15 mm silicon, a second layer of 20 mm silicon, a first electrode pair (“Filter1”) of 0.1 mm copper, and no second electrode pair (“Filter2”). Figure 18C shows a detection plot for a second variant with channel 4n+2, a first layer of 30 mm silicon, a second layer of 20 mm silicon, a first electrode pair (“Filter1”) of 0.1 mm copper, and no second electrode pair (“Filter2”). Figure 18D shows a detection plot for a third variant with channel 4n+3, a first layer of 25 mm silicon, a second layer of 50 mm silicon, a first electrode pair (“Filter1”) of 0.2 mm copper, and a second electrode pair ("Filter2") of 0.2 mm copper. In each plot, the y-axis is photons/second and the x-axis is energy (keV). In each, the first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot. Referring to Figures 18A-18D, it can be seen that providing the electrodes leads to improved detection.

EXAMPLE 4

A detector utilizing voltage-controlled layers as described herein was constructed and used to detect a Gd contrast agent in comparison with detection using a photon-counting detector with 10 keV spectrum resolution, four energy bins, and 1440 projections per turn.

Figure 19 shows a decomposed image of a Gd contrast agent. Figure 20 shows a detection plot at varying energies based on the inset equation. The y-axis of the detection plot is photons/second. Figures 21A, 21B, and 21C show enlarged versions of the water basis component, the bone basis component, and the Gd basis component, respectively, of the image in Figure 19. The image can be decomposed based on: $\mu = (a_1)(\mu_{water}) + (a_2)(\mu_{bone}) + (a_3)(\mu_{Gd})$. The image can be reconstructed using iterative maximum-likelihood and prior image constraint based on:

$$\max_{\mu} \sum_{i} \left(I_i \cdot \ln\left(\hat{I}_i\right) - \hat{I}_i \right) - \left(\mu - \mu'\right)^{T} \cdot \overset{\text{Weighting Factor}}{\Gamma} \cdot \left(\mu - \underset{\text{Dual-material Decomposition}}{\mu'}\right)$$

Figures 22A, 22B, 22C, and 22D show images of Gd contrast agent (top portion in each figure) at a concentration of 0.06 mol/L, 0.09 mol/L, 0.12 mol/L, and 0.15 mol/L, respectively, and a reconstruction of that image (bottom portion).

Figure 23 shows a detection plot at varying Gd concentration for components of an image of Gd contrast dye for both the system of the subject invention (circle data points, the line that is higher in the plot) and the photon-counting detector (square data points, the line that is lower in the plot). Referring to Figure 23, the system of the subject invention displays good detection performance.

EXAMPLE 5

A ray model was investigated for spectral sampling. Figure 24A shows a schematic view of a radiation source providing radiation through a human subject to a detector, according to the ray model. Figure 24B shows two detection plots at varying energy for the ray model. The y-axis of each detection plot is photons/second. Figure 25 shows a detection plot of the source character at varying energy after 22 cm of water attenuation. The ray model after linearization can be summarized as follows (the lower the condition number, the better the reconstruction):

$$I(v,d) = I_0 \int_0^{E_{max}} S(E,v)D(E,d)\exp\left\{-\int_{-\infty}^{\infty} \mu(E,r)\,dr\right\} dE$$

$$= I_0 \int_0^{E_{max}} S(E,v)D(E,d)\exp\left\{-\int_{-\infty}^{\infty} [\mu_0(E,r) + \Delta\mu(E,r)]\,dr\right\} dE$$

$$= I_0 \int_0^{E_{max}} S(E,v)D(E,d)\exp\left\{-\int_{-\infty}^{\infty} \mu_0(E,r)\,dr\right\}\left(1 - \int_{-\infty}^{\infty} \Delta\mu(E,r)\,dr\right) dE$$

Discretization

$$A\Delta\mu = b$$

Figures 26A, 26B, 26C, and 26D show detection plots for first and second layers of an attenuated detector at varying percentages of first layer/entire detector substrate and second layer/entire detector substrate. In each plot, the first layer is closer to the X-ray source during detection, and has the higher value at lower energy in the plot. Figure 26A is for a detector substrate of 30 mm silicon with percentages (first layer/entire detector substrate, second layer/entire detector substrate) of 25%, 75% and shows a correlation of 0.9576. Figure 26B is for a detector substrate of 30 mm silicon with percentages (first layer/entire detector substrate, second layer/entire detector substrate) of 35%, 65% and shows a correlation of 0.9585. Figure 26C is for a detector substrate of 30 mm silicon with percentages (first layer/entire detector substrate, second layer/entire detector substrate) of 45%, 55% and shows a correlation of 0.9593. Figure 26D is for a detector substrate of 30 mm silicon with percentages (first layer/entire detector substrate, second layer/entire detector substrate) of 55%, 45% and shows a correlation of 0.9598.

EXAMPLE 6

A detector utilizing three voltage-controlled layers (triple-layer detector) as described herein was modeled using the ray model discussed in Example 5. The detector was 0.15 cm CdTe, with layer depths of 0.019 cm, 0.170 cm, and 0.052 cm. The condition number was 8.19. Figure 27A shows a detection plot for the first, second, and third layers in this three-layer detector. This detector showed good detection performance.

EXAMPLE 7

A detector utilizing three voltage-controlled layers (triple-layer detector) as described herein was modeled using the ray model discussed in Example 5, with attenuation. The detector was 0.15 cm CdTe. The range was up to 140 keV after 30 cm of water attenuation. The best voltages were 83 kVp, 114 kVp, and 140 kVp, and the condition number was 19.72. Figure 27B shows a detection plot for the first, second, and third layers in this three-layer detector. This detector showed good detection performance.

EXAMPLE 8

A detector utilizing three voltage-controlled layers (triple-layer detector) as described herein was modeled using a ray model by basis materials. The detector was 0.15 cm CdTe. The ray model by basis materials can be summarized as follows:

$$I(v,d) = I_0 \int_0^{E_{max}} S(E,v)D(E,d)\exp\left\{-\int_{-\infty}^{\infty} \mu(E,r)dr\right\}dE$$

$$I_m(v,d) - I(v,d) = I_0 \int_0^{E_{max}} S(E,v)D(E,d)\boldsymbol{B}(E)dE \cdot \Delta\boldsymbol{a}$$

$$\boldsymbol{B}(E) = \left[\mu_{water}(E), \mu_{bone}(E), \mu_{Gd}(E)\right]$$

$$\Delta\boldsymbol{a} = \left[\int_{-\infty}^{\infty} \Delta a_1(r)dr, \int_{-\infty}^{\infty} \Delta a_2(r)dr, \int_{-\infty}^{\infty} \Delta a_3(r)dr\right]$$

The layer depths of the detector were 0.0067 cm, 0.072 cm, and 0.011 cm. The condition number was 61871. Figure 27C shows a detection plot for the first, second, and third layers in this three-layer detector. This detector showed good detection performance.

EXAMPLE 9

A detector utilizing three voltage-controlled layers (triple-layer detector) as described herein was modeled using the ray model discussed in Example 8, with attenuation. The detector was 0.15 cm CdTe. The range was up to 140 keV after 30 cm of water attenuation. The best voltages were 70 kVp, 108 kVp, and 140 kVp, and the condition number was 76794. Figure 27D shows a detection plot for the first, second, and third layers in this three-layer detector. This detector showed good detection performance.

It should be understood that the examples and embodiments described herein are for illustrative purposes only and that various modifications or changes in light thereof will be suggested to persons skilled in the art and are to be included within the spirit and purview of this application.

All patents, patent applications, provisional applications, and publications referred to or cited herein (including those in the “References” section) are incorporated by reference in their entirety, including all figures and tables, to the extent they are not inconsistent with the explicit teachings of this specification.

REFERENCES

Overdick, Michael; Baumer, Christian; Engel, K.J.; et al, "Status of Direct Conversion Detectors for Medical Imaging With X-Rays," in Nuclear Science, IEEE Transactions on , vol.56, no.4, pp.1800-1809, Aug. 2009.

Doran S J, Koerkamp K K, Bero M A, et al. A CCD-based optical CT scanner for high-resolution 3D imaging of radiation dose distributions: equipment specifications, optical simulations and preliminary results[J]. Physics in medicine and biology, 2001, 46(12): 3191.

Pan D, Roessl E, Schlomka J P, et al. Computed Tomography in Color: NanoK-Enhanced Spectral CT Molecular Imaging[J]. Angewandte Chemie, 2010, 122(50): 9829-9833.

Chu J, Cong W, Li L, et al. Combination of current-integrating/photon-counting detector modules for spectral CT[J]. Physics in medicine and biology, 2013, 58(19): 7009.

Taguchi K, Iwanczyk J S. Vision 20/20: Single photon counting x-ray detectors in medical imaging[J]. Medical physics, 2013, 40(10): 100901.

Bornefalk H, Danielsson M. Photon-counting spectral computed tomography using silicon strip detectors: a feasibility study[J]. Physics in medicine and biology, 2010, 55(7): 1999.

Persson M, Huber B, Karlsson S, et al. Energy-resolved CT imaging with a photon-counting silicon-strip detector[J]. Physics in medicine and biology, 2014, 59(22): 6709.

Gruner S M, Tate M W, Eikenberry E F. Charge-coupled device area X-ray detectors[J]. Review of Scientific Instruments, 2002, 73(8): 2815-2842.

Alvarez R E, Macovski A. Energy-selective reconstructions in x-ray computerised tomography[J]. Physics in medicine and biology, 1976, 21(5): 733.

Shikhaliev P M. Projection x-ray imaging with photon energy weighting: experimental evaluation with a prototype detector[J]. Physics in medicine and biology, 2009, 54(16): 4971.

Giersch J, Niederlöhner D, Anton G. The influence of energy weighting on X-ray imaging quality[J]. Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment, 2004, 531(1): 68-74.

Shikhaliev P M. Energy-resolved computed tomography: first experimental results[J]. Physics in medicine and biology, 2008, 53(20): 5595.

Taguchi K, Frey E C, Wang X, et al. An analytical model of the effects of pulse pileup on the energy spectrum recorded by energy resolved photon counting x-ray detectors[J]. Medical physics, 2010, 37(8): 3957-3969.

Burke, B.E.; Mountain, R.W.; Daniels, P.J.; et al, "CCD soft X-ray imaging spectrometer for the ASCA satellite," in Nuclear Science, IEEE Transactions on , vol.41, no.1, pp.375-385, Feb 1994.

Lundqvist, M.; Cederstrom, B.; Chmill, V.; et al, "Computer simulations and performance measurements on a silicon strip detector for edge-on imaging," in Nuclear Science, IEEE Transactions on , vol.47, no.4, pp.1487-1492, Aug 2000.

Bertolini G, Coche A. SEMICONDUCTOR DETECTORS[J]. 1968.

Marcelot, O.; Estribeau, M.; Goiffon, V.; et al, "Study of CCD Transport on CMOS Imaging Technology: Comparison Between SCCD and BCCD, and Ramp Effect on the CTI," in Electron Devices, IEEE Transactions on , vol.61, no.3, pp.844-849, March 2014.

Tompsett, M.F., "Surface potential equilibration method of setting charge in charge-coupled devices," in Electron Devices, IEEE Transactions on , vol.22, no.6, pp.305-309, Jun 1975.

Hoople, C.R.; Krusius, J.P., "Characteristics of submicrometer gaps in buried-channel CCD structures," in Electron Devices, IEEE Transactions on , vol.38, no.5, pp.1175-1181, May 1991.

Arfelli, F.; Barbiellini, G.; Bonvicini, V.; et al, "An “edge-on” silicon strip detector for X-ray imaging," in Nuclear Science, IEEE Transactions on , vol.44, no.3, pp.874-880, Jun 1997.

L. Rigon, F. Arfelli, A. Astolfo, et al. A single-photon counting “edge-on” silicon detector for synchrotron radiation mammography, Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment, Volume 608, Issue 1, Supplement, 1 September 2009, Pages S62-S65.

International Patent Application No. PCT/US2015/067441, filed December 22, 2015.

U.S. Provisional Application Serial No. 62/095,235, filed December 22, 2014.

Q. Xu et al., Image Reconstruction for Hybrid True-Color Micro-CT, IEEE Transactions on Biomedical Engineering, Vol. 50, No. 6, June 2012.

B. De Man et al., An Iterative Maximum-Likelihood Polychromatic Algorithm for CT, IEEE Transactions on Medical Imaging, Vol. 20, No. 10, October 2001.

Zou, U.S. Patent Application Publication No. 2013/0251097, published September 26, 2013.

D. Gierada et al., Gadolinium as a CT contrast agent: assessment in a porcine model, Radiology 210:829-834, 1999.

CLAIMS

What is claimed is:

1. An imaging system, comprising:

a computed tomography (CT) scanner including an X-ray source; and

a detector for receiving X-ray radiation from the X-ray source after it passes through a sample to be imaged,

wherein the detector includes a first pair of electrodes and a second pair of electrodes disposed thereon and configured to provide a first voltage and a second voltage, respectively, to the detector.

2. The imaging system according to claim 1, wherein the detector includes a first layer and a second layer,

wherein the first pair of electrodes is disposed on the first layer and applies the first voltage to the first layer, and

wherein the second pair of electrodes is disposed on the second layer and applies the second voltage to the second layer.

3. The imaging system according to any of claims 1-2, wherein the detector comprises at least one of material selected from cadmium zinc telluride (CZT), silicon, and cadmium telluride (CdTe).

4. The imaging system according to any of claims 1-3, wherein the first voltage is different from the second voltage.

5. The imaging system according to any of claims 2-4, wherein the first layer and the second layer are defined by the first and second pairs of electrodes, respectively.

6. The imaging system according to any of claims 1-5, wherein both electrodes of the first pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

7. The imaging system according to any of claims 1-6, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector not having any of the first pair of electrodes disposed thereon.

8. The imaging system according to any of claims 1-7, wherein both electrodes of the second pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

9. The imaging system according to any of claims 1-8, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector not having any of the second pair of electrodes disposed thereon.

10. The imaging system according to any of claims 1-5 and 7-9, wherein both electrodes of the first pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

11. The imaging system according to any of claims 1-6 and 8-10, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector having both electrodes of the first pair of electrodes disposed thereon.

12. The imaging system according to any of claims 1-7 and 9-11, wherein both electrodes of the second pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

13. The imaging system according to any of claims 1-8 and 9-12, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector having both electrodes of the second pair of electrodes disposed thereon.

14. The imaging system according to any of claims 2-13, wherein the detector further comprises:

a third pair of electrodes disposed thereon and configured to provide a third voltage to the detector;

a fourth pair of electrodes disposed thereon and configured to provide a fourth voltage to the detector;

a third layer; and

a fourth layer,

wherein the first to fourth pairs of electrodes are disposed on and configured to provide the first to fourth voltages to the first to fourth layers, respectively.

15. The imaging system according to claim 14, wherein at least one of the first to fourth voltages is different from at least one of the other voltages of the first to fourth voltages.

16. The imaging system according to any of claims 14-15, wherein the first, second, third, and fourth layers are defined by the first to fourth pairs of electrodes, respectively.

17. The imaging system according to any of claims 1-16, further comprising a controller configured to control the pairs of electrodes (and third and fourth pairs of electrodes, if present) to apply the first and second voltages (and the third and fourth voltages, if the third and fourth pairs of electrodes are present) such that photons in the detector are captured in a high energy bin and a low energy bin.

18. The imaging system according to claim 17, wherein the controller is configured to control the pairs of electrodes to apply the voltages such that photons in the detector are captured in the high energy bin and the low energy bin in a predetermined ratio.

19. The imaging system according to any of claims 2-18, wherein a thickness of each layer of the detector present is the same as that of each other layer of the detector present.

20. The imaging system according to any of claims 2-18, wherein a thickness of at least one layer of the detector present is different from that of at least one other layer of the detector present.

21. The imaging system according to any of claims 2-18, wherein a thickness of each layer of the detector present is the different from that of each other layer of the detector present.

22. The imaging system according to any of claims 1-21, further comprising a machine-readable medium having machine-executable instructions for performing an energy resolving process on the collected X-ray radiation.

23. The imaging system according to claim 22, wherein the energy resolving process includes:

determining the generated charge density within the detector using Formula 1:

Formula 1 - $m_1E_1N_1a_{k1} + m_2E_2N_2a_{k2} + \cdots + m_nE_nN_na_{kn} = g_k(x)$,

where E_i is photon energy, N_i is photon density with energy E_i, a_{ki} is the attenuation coefficient of photon with energy E_i for the given material thickness, m_i is an empirical coefficient that represents the number of generated charges by photons with energy E_i per energy unit, and $g_k(x)$ is the generated charge density within the material of the detector of a specific thickness; and

repeating the determination of the generated charge density at a different thickness within the material of the detector.

24. The imaging system according to any of claims 22-23, further comprising a processer, wherein the energy resolving process is performed by the processor.

25. The imaging system according to any of claims 1-24, configured to perform imaging on a sample that is a part of a human patient.

26. The imaging system according to any of claims 1-25, wherein the X-ray source is configured to provide X-ray radiation that has an energy of from 10 keV to 120 keV.

27. The imaging system according to any of claims 1-25, wherein the X-ray source is configured to provide X-ray radiation that has an energy of less than 20 keV.

28. The imaging system according to any of claims 1-25, wherein the X-ray source is configured to provide X-ray radiation that has an energy of more than 20 keV.

29. The imaging system according to any of claims 1-28, wherein the CT scanner has third-generation geometry.

30. The imaging system according to any of claims 1-28, wherein the CT scanner has fourth-generation geometry.

31. A method of imaging, comprising:
providing the imaging system according to any of claims 1-30; and
using the imaging system for its intended purpose to image the sample.

32. A method of imaging, comprising:
providing X-ray radiation to a sample to be imaged;
collecting the X-ray radiation with a detector;
providing a first voltage to the detector using a first pair of electrodes disposed thereon; and
providing a second voltage to the detector using a second pair of electrodes disposed thereon.

33. The method according to claim 32, wherein the detector includes a first layer and a second layer.

34. The method according to claim 33, wherein the first pair of electrodes is disposed on the first layer and applies the first voltage to the first layer, and

wherein the second pair of electrodes is disposed on the second layer and applies the second voltage to the second layer.

35. The method according to any of claims 32-34, wherein the detector comprises at least one of material selected from cadmium zinc telluride (CZT), silicon, and cadmium telluride (CdTe).

36. The method according to any of claims 32-35, wherein the first voltage is different from the second voltage.

37. The method according to any of claims 32-36, further comprising dynamically modulating at least one of the first voltage and the second voltage such that photons in the detector are captured in a high energy bin and a low energy bin in a predetermined ratio.

38. The method according to any of claims 32-37, further comprising dynamically modulating both the first voltage and the second voltage such that photons in the detector are captured in a high energy bin and a low energy bin in a predetermined ratio.

39. The method according to any of claims 37-38, wherein the ratio (higher energy bin : lower energy bin) is 2:1, 3:1, 4:1, 1:2, 1:3, 1:4, 5:1, 1:5, 1.5:1, 1:1.5, 2.5:1, 1:2.5, 3.5:1, 1:3.5, 4.5:1, or 1:4.5.

40. The method according to any of claims 32-39, further comprising dividing the X-ray spectrum into low and high energies at different proportions during detection such that detected X-ray radiation is classified as either high energy or low energy during detection by modulating the first and second voltages.

41. The method according to claim 40, wherein the low and high energies are divided into proportions of (low/high) 10%/90%, 20%/80%, 70%/30%, 60%/40%, 50%/50%, 40%/60%, 30%,70%, 20%/80%, and 10%/90% during detection by modulating the first and second voltages.

42. The method according to any of claims 40-41, further comprising distributing these detector variants in a natural sequence, and repeating until a full detector ring is covered in third generation geometry.

43. The method according to any of claims 40-41, further comprising distributing these detector variants in a natural sequence, and repeating until a full detector ring is covered in fourth generation geometry.

44. The method according to any of claims 33-43, wherein the first layer and the second layer are defined by the first and second pairs of electrodes, respectively.

45. The method according to any of claims 32-44, wherein both electrodes of the first pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

46. The method according to any of claims 32-45, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector not having any of the first pair of electrodes disposed thereon.

47. The method according to any of claims 32-46, wherein both electrodes of the second pair of electrodes are disposed parallel to a direction of X-rays incoming from the X-ray source.

48. The method according to any of claims 32-48, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector not having any of the second pair of electrodes disposed thereon.

49. The method according to any of claims 32-44 and 46-48, wherein both electrodes of the first pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

50. The method according to any of claims 32-45 and 47-49, wherein both electrodes of the first pair of electrodes are disposed such that incoming X-rays from the X-ray source are first incident on a side of the detector having both electrodes of the first pair of electrodes disposed thereon.

51. The method according to any of claims 32-46 and 48-50, wherein both electrodes of the second pair of electrodes are disposed perpendicular to a direction of X-rays incoming from the X-ray source.

52. The method according to any of claims 32-47 and 49-51, wherein both electrodes of the second pair of electrodes are disposed such that incoming X-rays from the X-ray source are second incident on a side of the detector having both electrodes of the second pair of electrodes disposed thereon.

53. The method according to any of claims 32-52, wherein the detector further comprises:

a third pair of electrodes disposed thereon and configured to provide a third voltage to the detector;

a fourth pair of electrodes disposed thereon and configured to provide a fourth voltage to the detector;

a third layer; and

a fourth layer,

wherein the first to fourth pairs of electrodes are disposed on and configured to provide the first to fourth voltages to the first to fourth layers, respectively.

54. The method according to claim 53, wherein at least one of the first to fourth voltages is different from at least one of the other voltages of the first to fourth voltages.

55. The method according to any of claims 53-54, wherein the first, second, third, and fourth layers are defined by the first to fourth pairs of electrodes, respectively.

56. The method according to any of claims 32-55, further comprising controlling the first and second pairs of electrodes (and third and fourth pairs of electrodes, if present) to

apply the first and second voltages (and the third and fourth voltages, if the third and fourth pairs of electrodes are present) such that photons in the detector are captured in a high energy bin and a low energy bin.

57. The method according to claim 56, wherein the pairs of electrodes are controlled to apply the voltages such that photons in the detector are captured in the high energy bin and the low energy bin in a predetermined ratio.

58. The method according to any of claims 32-57, wherein a thickness of each layer of the detector present is the same as that of each other layer of the detector present.

59. The method according to any of claims 32-57, wherein a thickness of at least one layer of the detector present is different from that of at least one other layer of the detector present.

60. The method according to any of claims 32-57, wherein a thickness of each layer of the detector present is the different from that of each other layer of the detector present.

61. The method according to any of claims 32-60, further comprising performing an energy resolving process on the collected X-ray radiation.

62. The method according to claim 101, wherein the energy resolving process includes:

determining the generated charge density within the detector using Formula 1:

Formula 1 - $m_1E_1N_1a_{k1} + m_2E_2N_2a_{k2} + \cdots + m_nE_nN_na_{kn} = g_k(x)$,

where E_i is photon energy, N_i is photon density with energy E_i, a_{ki} is the attenuation coefficient of photon with energy E_i for the given material thickness, m_i is an empirical coefficient that represents the number of generated charges by photons with energy E_i per energy unit, and $g_k(x)$ is the generated charge density within the material of the detector of a specific thickness; and repeating the determination of the generated charge density at a different thickness within the material of the detector.

63. The method according to any of claims 61-62, wherein the energy resolving process is performed by a processor.

64. The method according to any of claims 32-63, wherein the sample to be imaged is a part of a human patient.

65. The method according to any of claims 32-64, wherein the X-ray radiation has an energy of from 10 keV to 120 keV.

66. The method according to any of claims 32-64, wherein the X-ray radiation has an energy of less than 20 keV.

67. The method according to any of claims 32-64, wherein the X-ray radiation has an energy of more than 20 keV.

68. The method according to any of claims 61-67, wherein the steps of the energy resolving process are stored on a machine-readable medium.

69. The method according to any of claims 32-68, wherein the imaging is a computed tomography (CT) scan.

70. The method according to any of claims 32-69, wherein the X-ray radiation is provided by an X-ray source of a CT scanner.

71. The method according to claim 70, wherein the CT scanner has third-generation geometry.

72. The method according to claim 70, wherein the CT scanner has fourth-generation geometry.

73. The method according to any of claims 32-72, wherein the detector is placed in an edge-on fashion during imaging, such that the X-ray irradiation enters a side of a substrate of the detector.

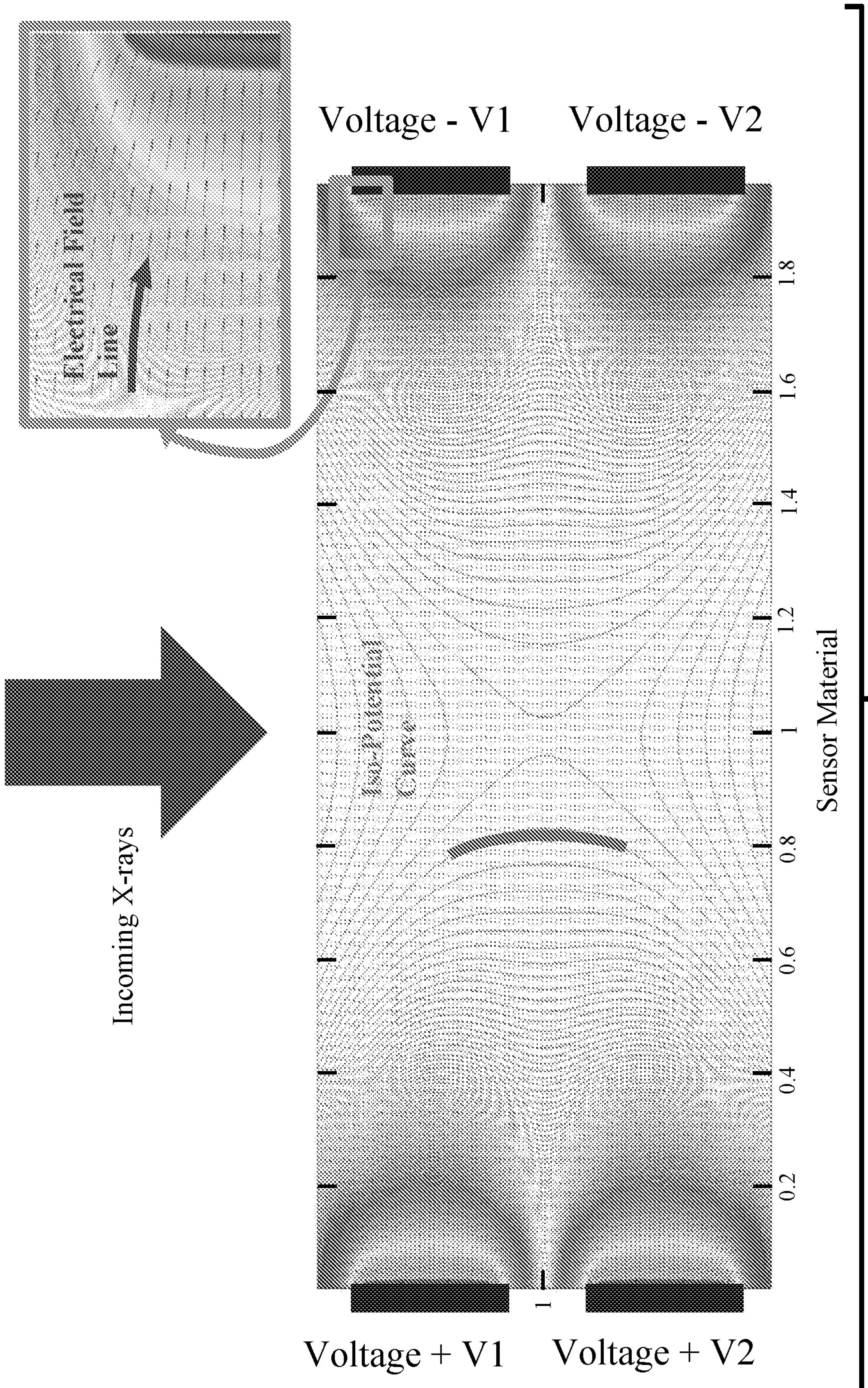

Voltage - V1
Voltage - V2
Voltage + V1
Voltage + V2
Electrical Field Line
Incoming X-rays
Iso-Potential Curve
Sensor Material
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8
1

FIG. 1A

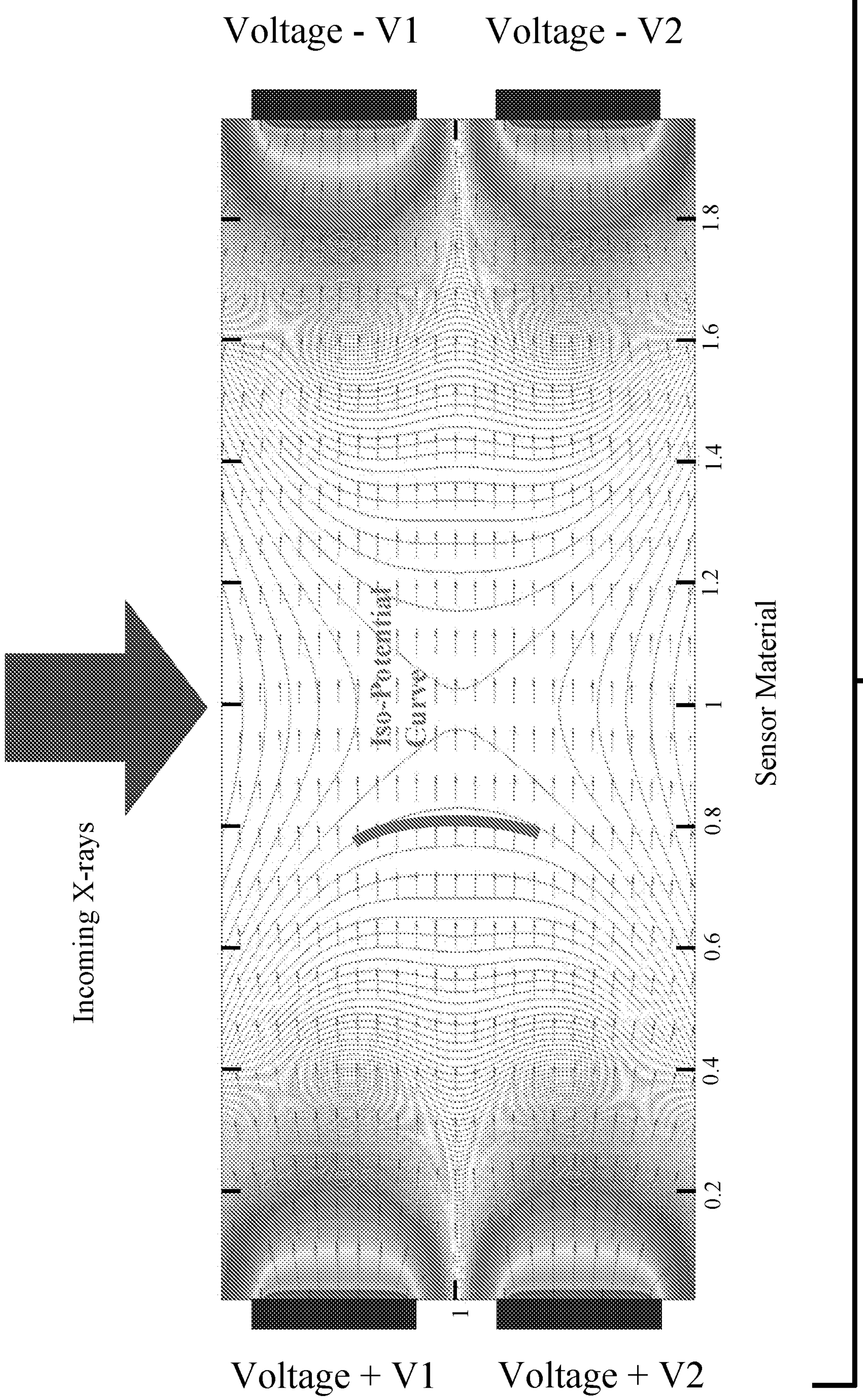

Voltage - V1
Voltage - V2
Incoming X-rays
Iso-Potential Curve
Sensor Material
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8
1
Voltage + V1
Voltage + V2

FIG. 1B

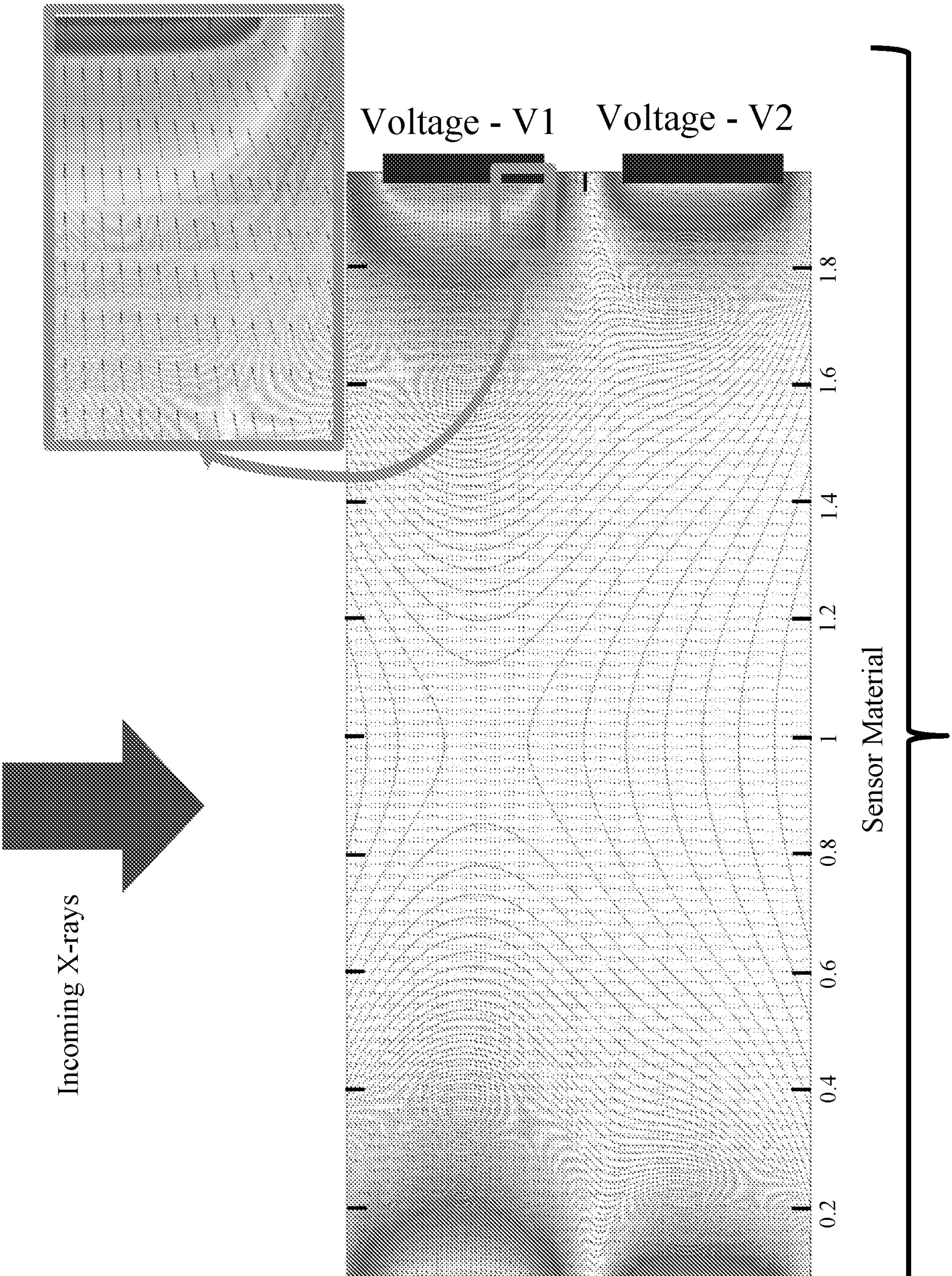

Voltage - V1
Voltage - V2
Voltage + V1
Voltage + V2
Incoming X-rays
Sensor Material
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8

FIG. 1C

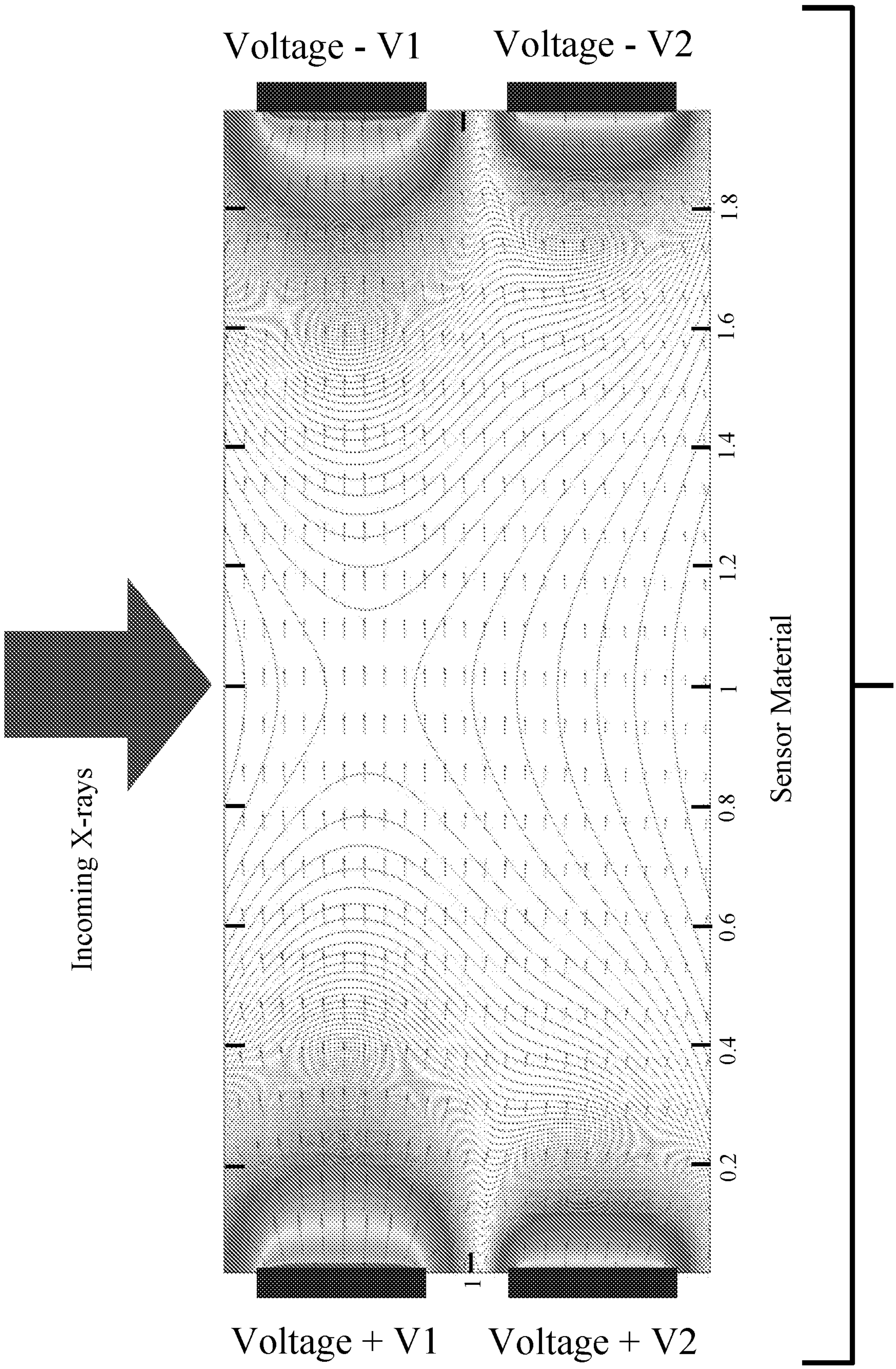

Voltage - V1
Voltage - V2
Incoming X-rays
Sensor Material
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8
Voltage + V1
Voltage + V2

FIG. 1D

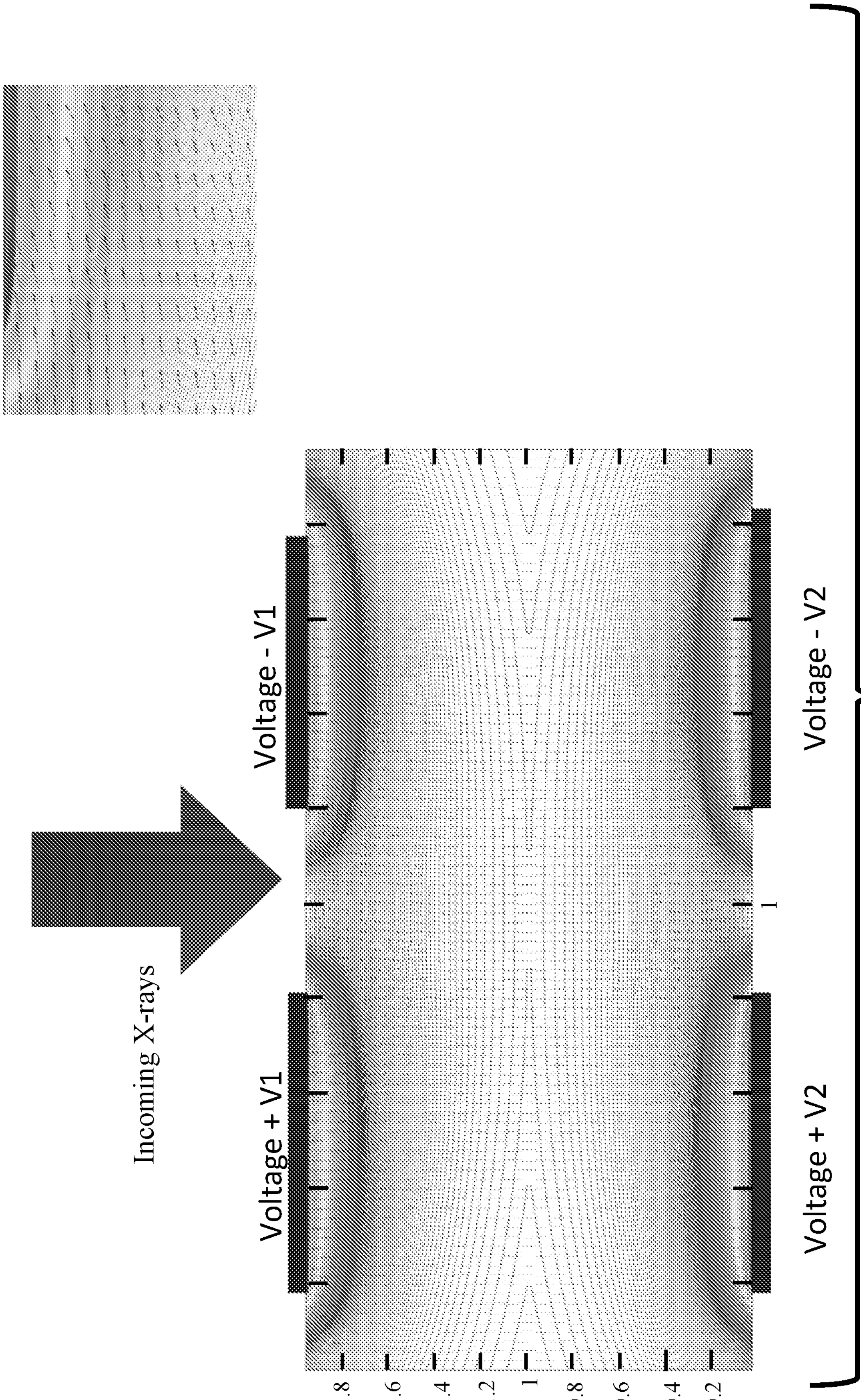

Incoming X-rays
Voltage + V1
Voltage - V1
Voltage + V2
Voltage - V2
1.8
1.6
1.4
1.2
1
0.8
0.6
0.4
0.2
1

FIG. 1E

FIG. 1F

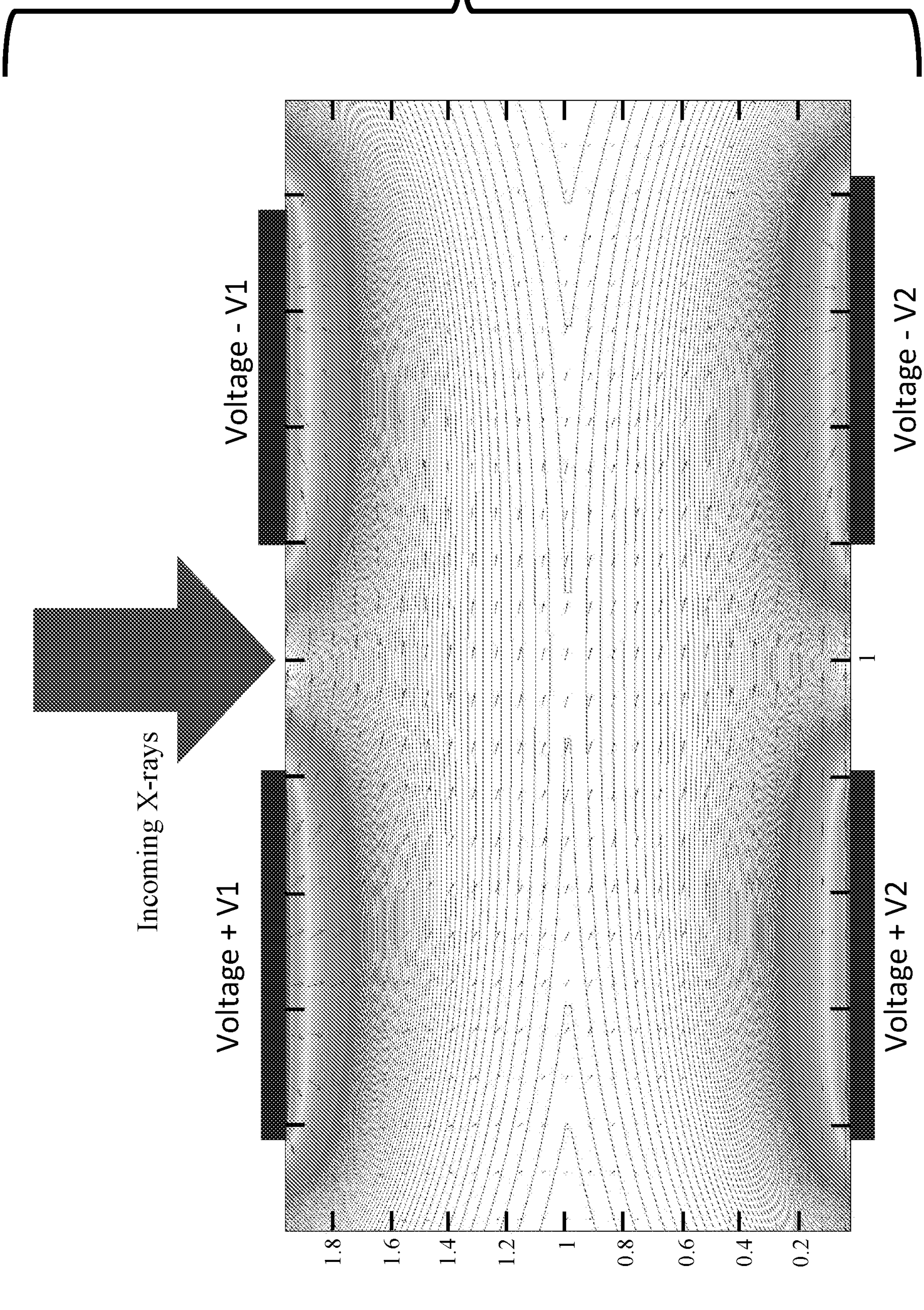

Voltage - V1
Voltage + V1
Incoming X-rays
Voltage - V2
Voltage + V2
1.8
1.6
1.4
1.2
1
0.8
0.6
0.4
0.2
1

FIG. 1G

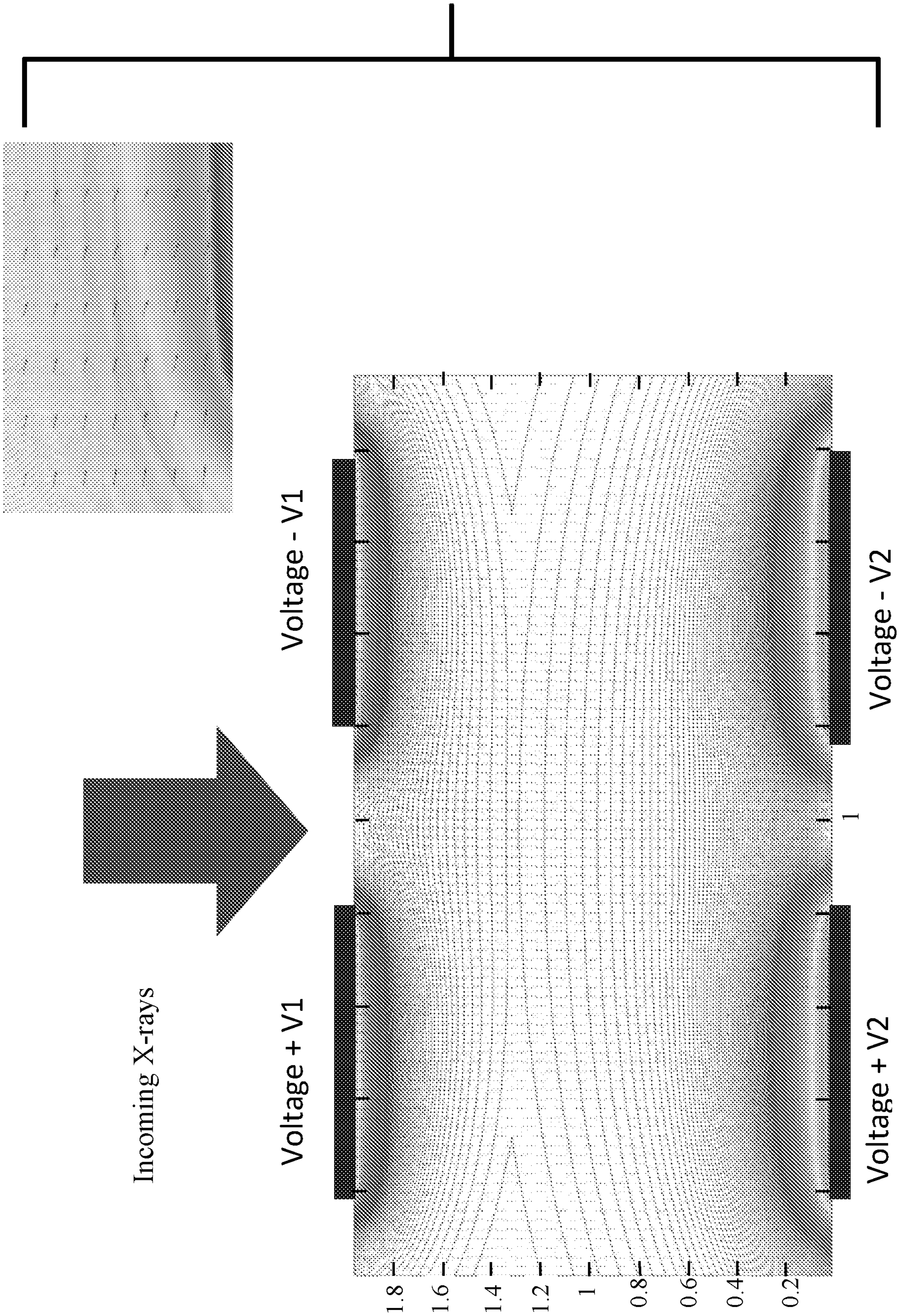
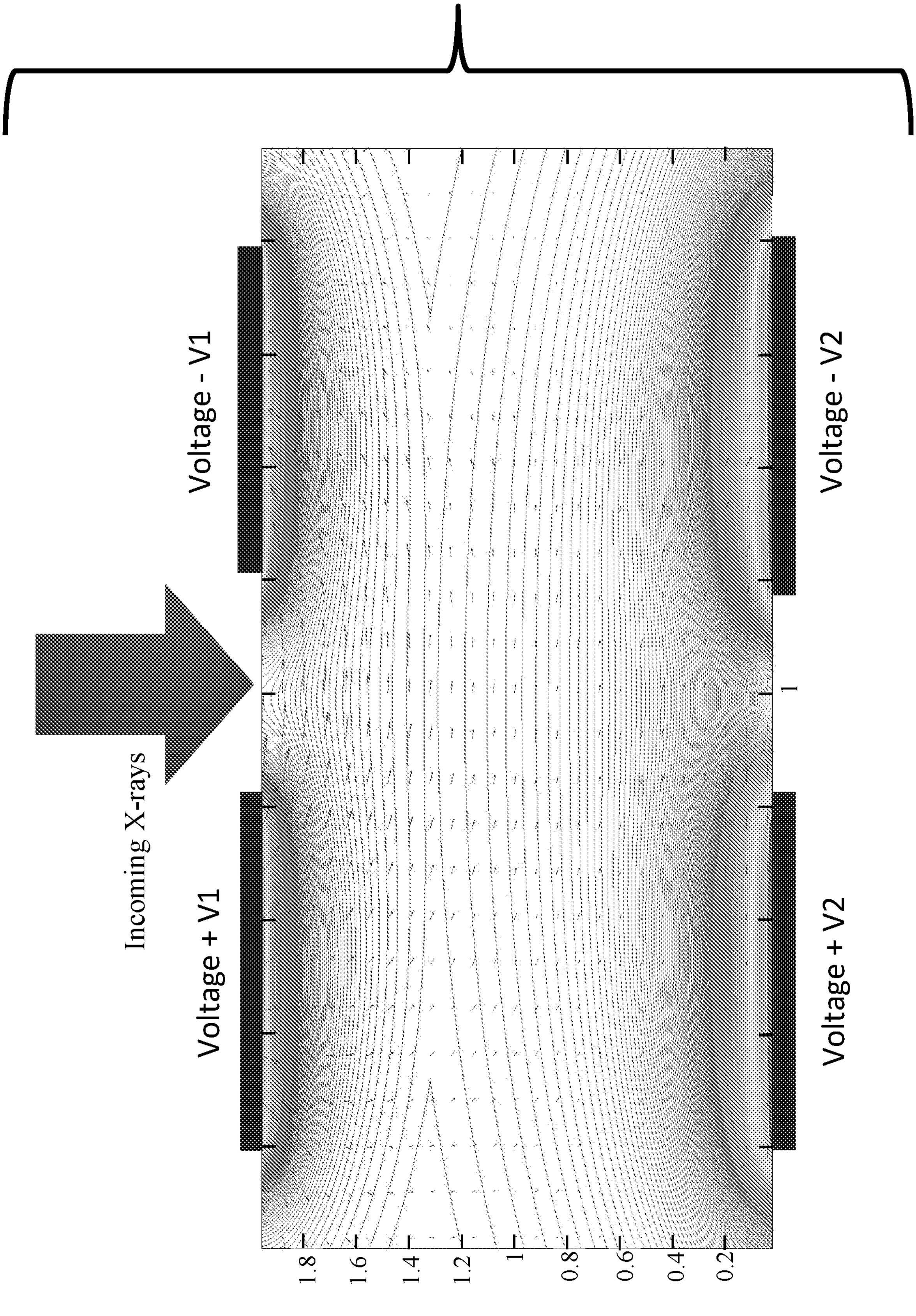

Incoming X-rays
Voltage + V1
Voltage - V1
Voltage + V2
Voltage - V2
1.8
1.6
1.4
1.2
1
0.8
0.6
0.4
0.2
1

FIG. 1H



SUBSTITUTE SHEET (RULE 26)

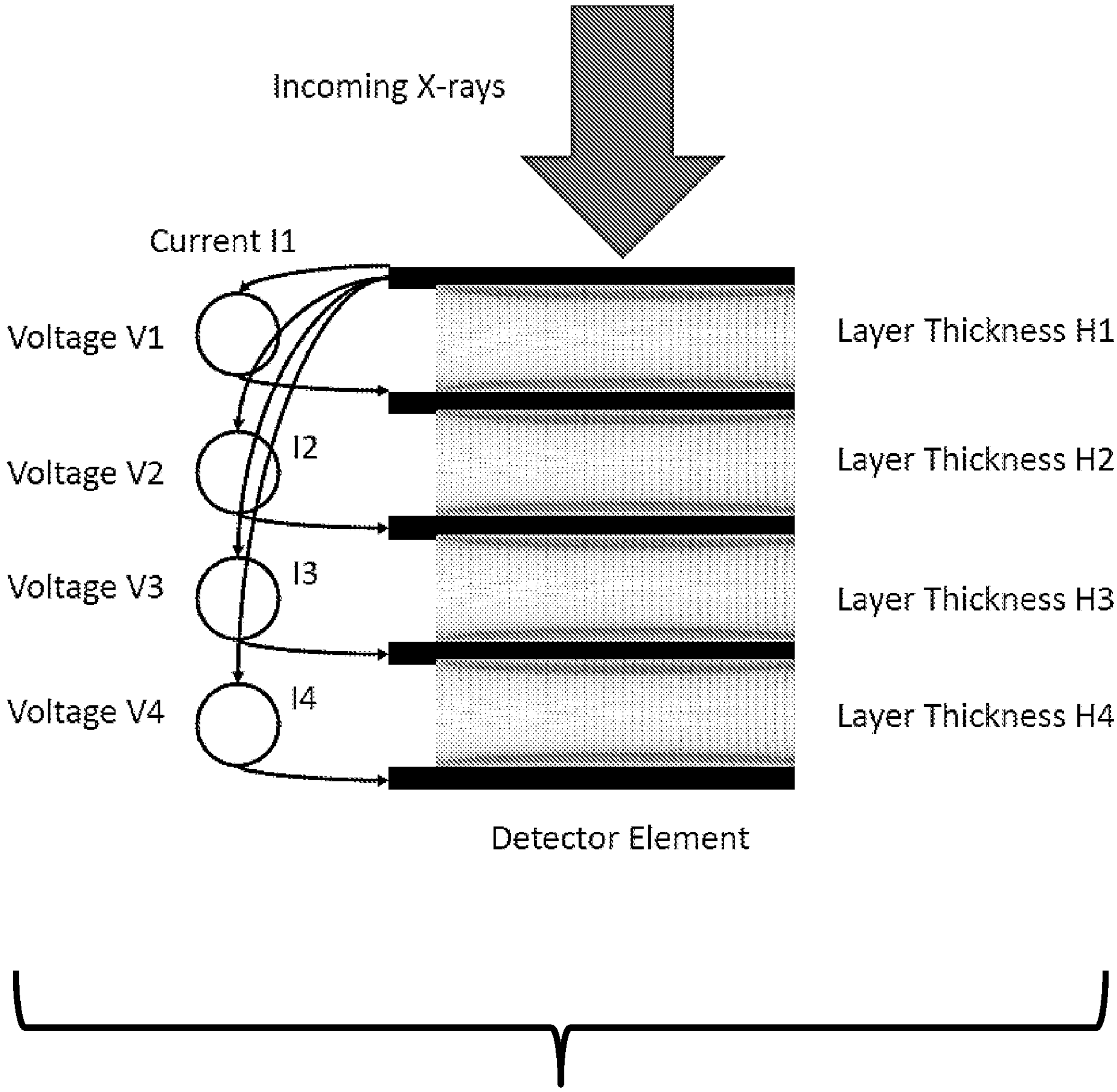

Incoming X-rays
Current I1
Voltage V1
Voltage V2
Voltage V3
Voltage V4
I2
I3
I4
Layer Thickness H1
Layer Thickness H2
Layer Thickness H3
Layer Thickness H4
Detector Element

FIG. 2

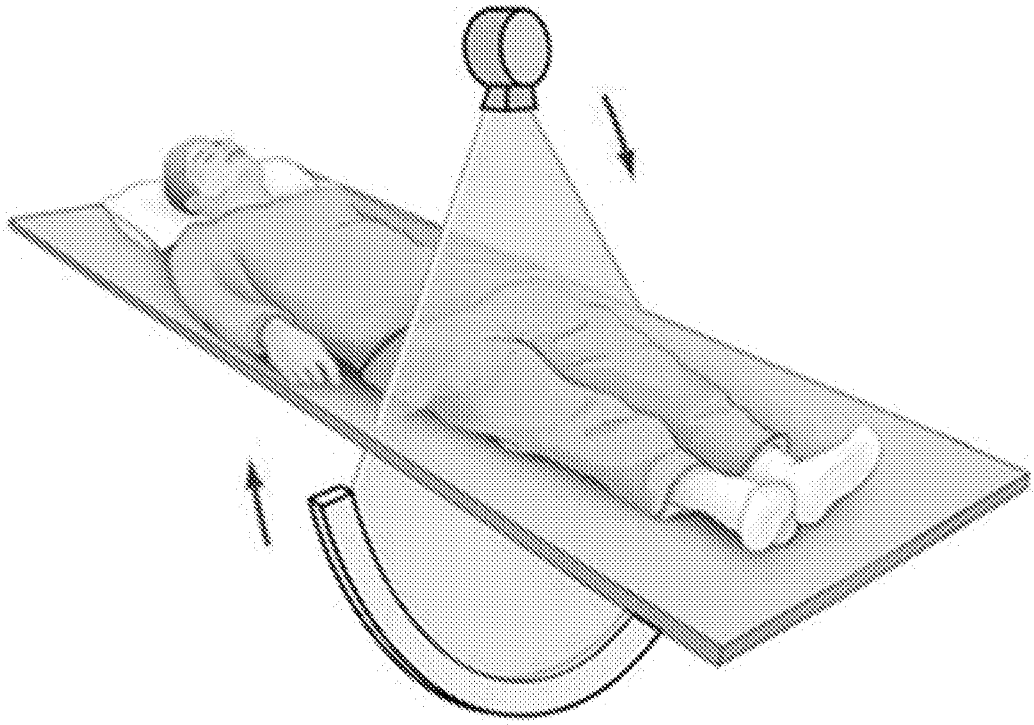


FIG. 3

FIG. 4A

FIG. 4B

FIG. 4C

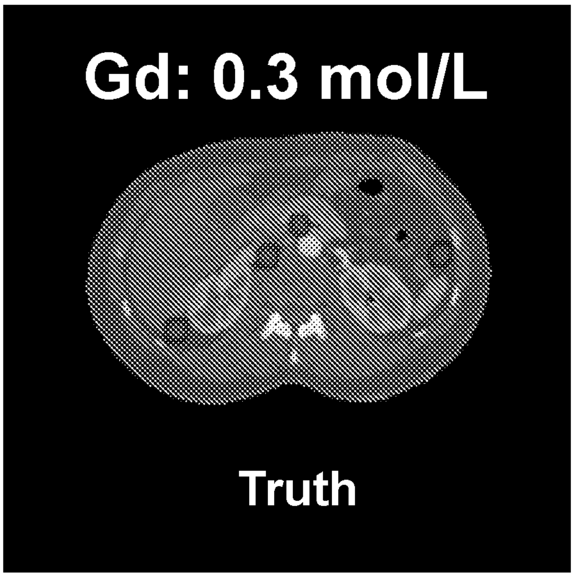


FIG. 5A

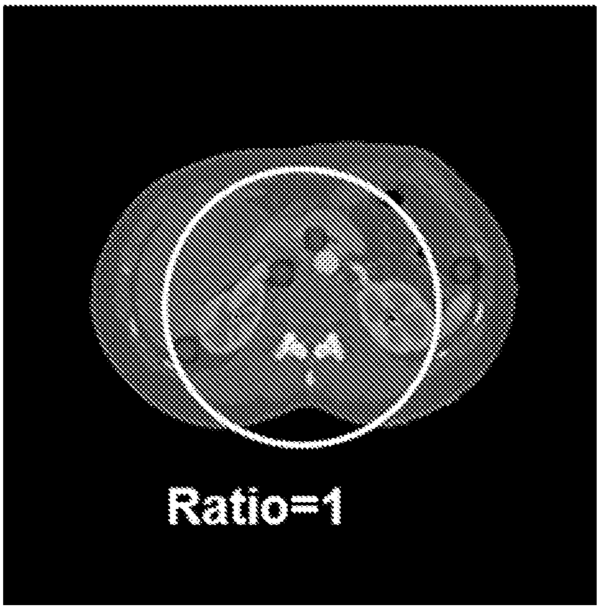


FIG. 5B

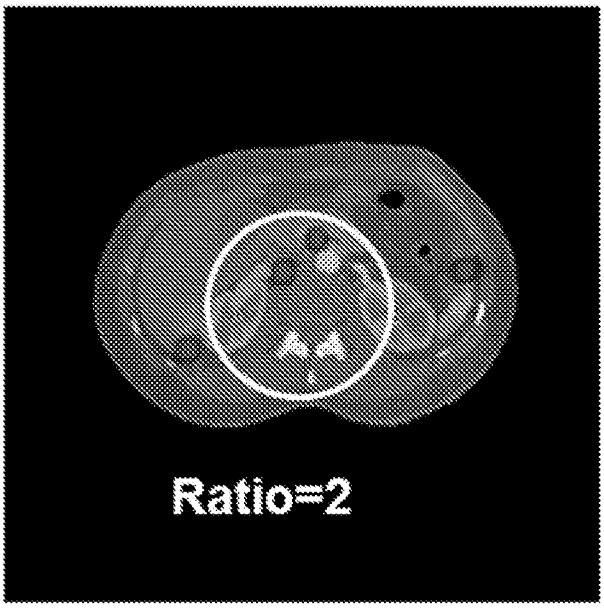


FIG. 5C

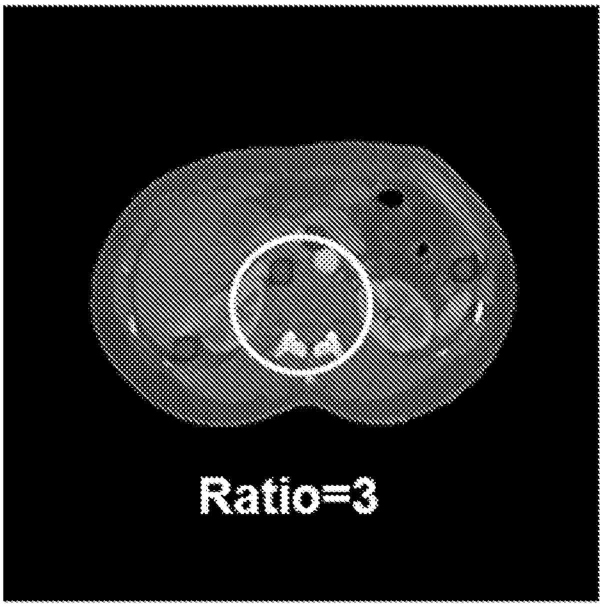


FIG. 5D

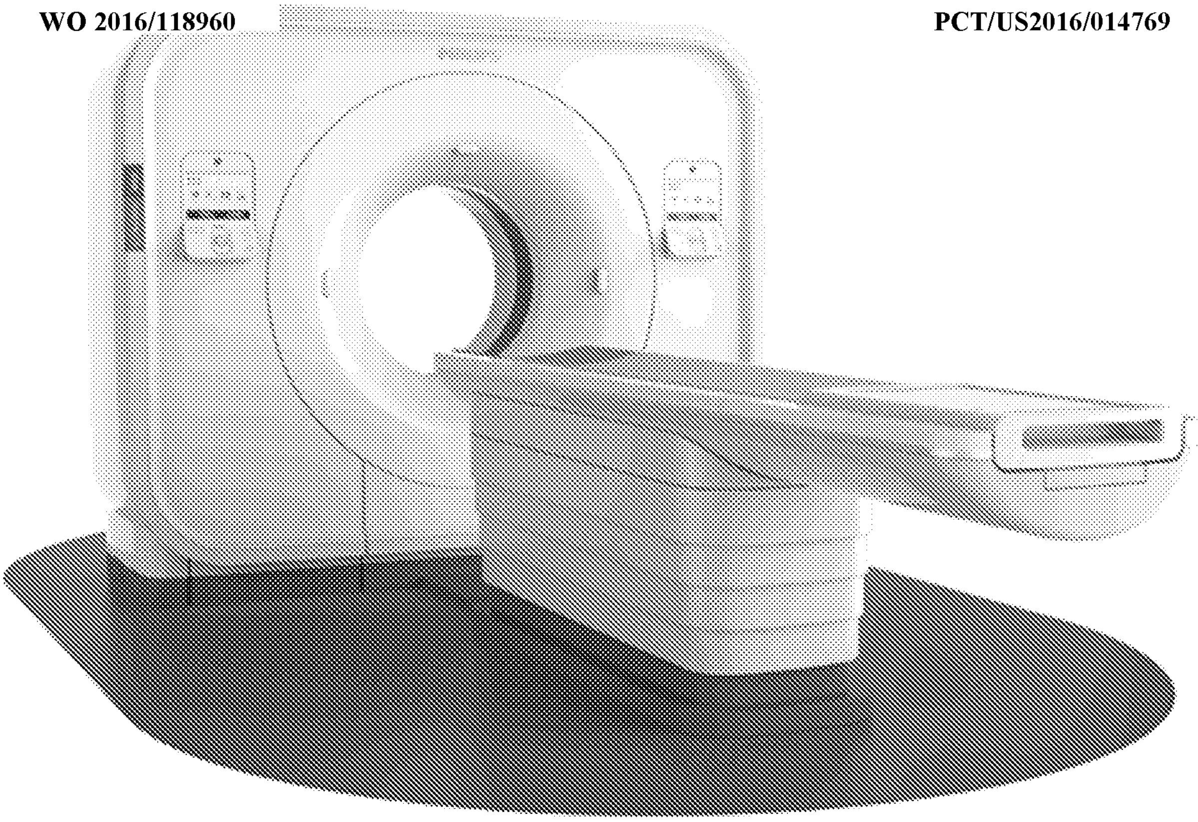


FIG. 6

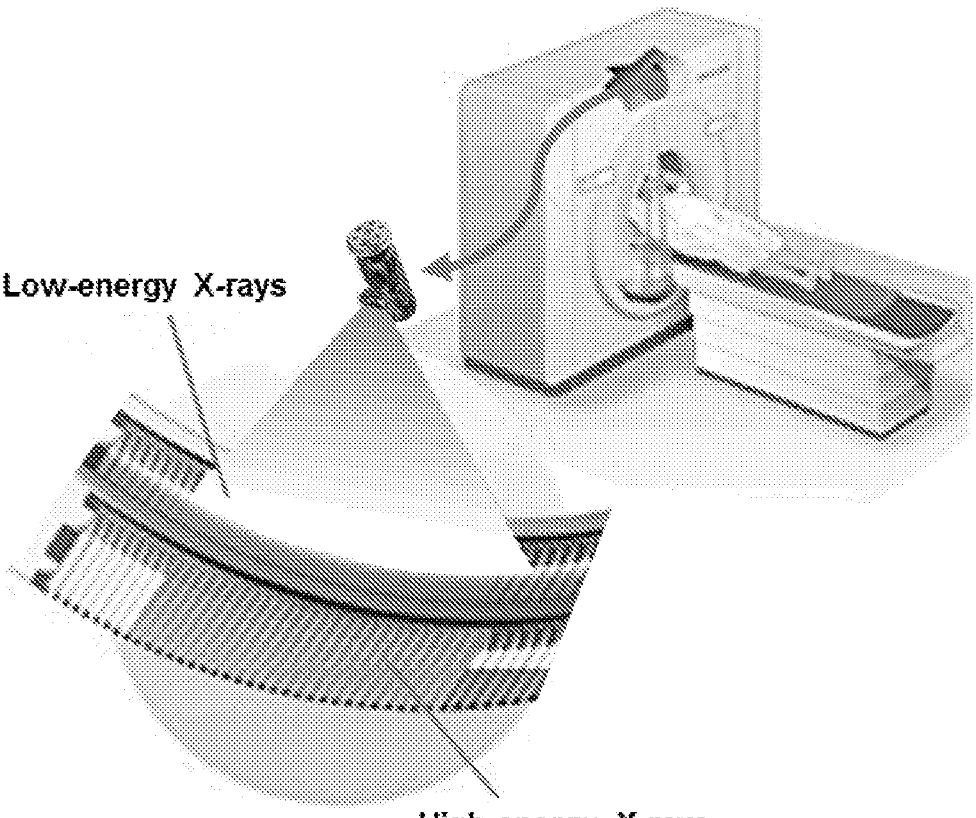


FIG. 7

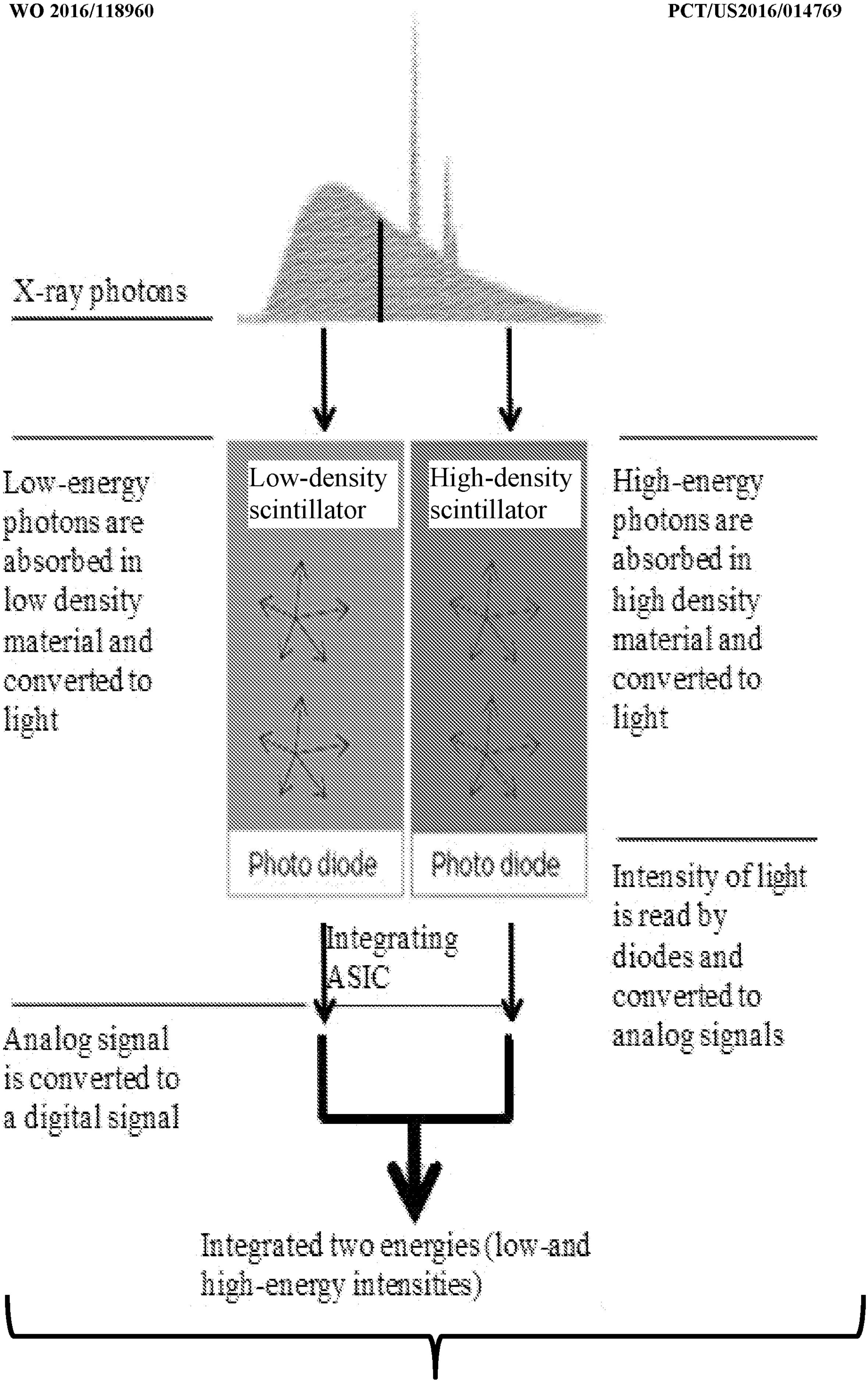

X-ray photons
Low-density scintillator
High-density scintillator
Low-energy photons are absorbed in low density material and converted to light
High-energy photons are absorbed in high density material and converted to light
Photo diode
Photo diode
Intensity of light is read by diodes and converted to analog signals
Integrating ASIC
Analog signal is converted to a digital signal
Integrated two energies (low-and high-energy intensities)

FIG. 8

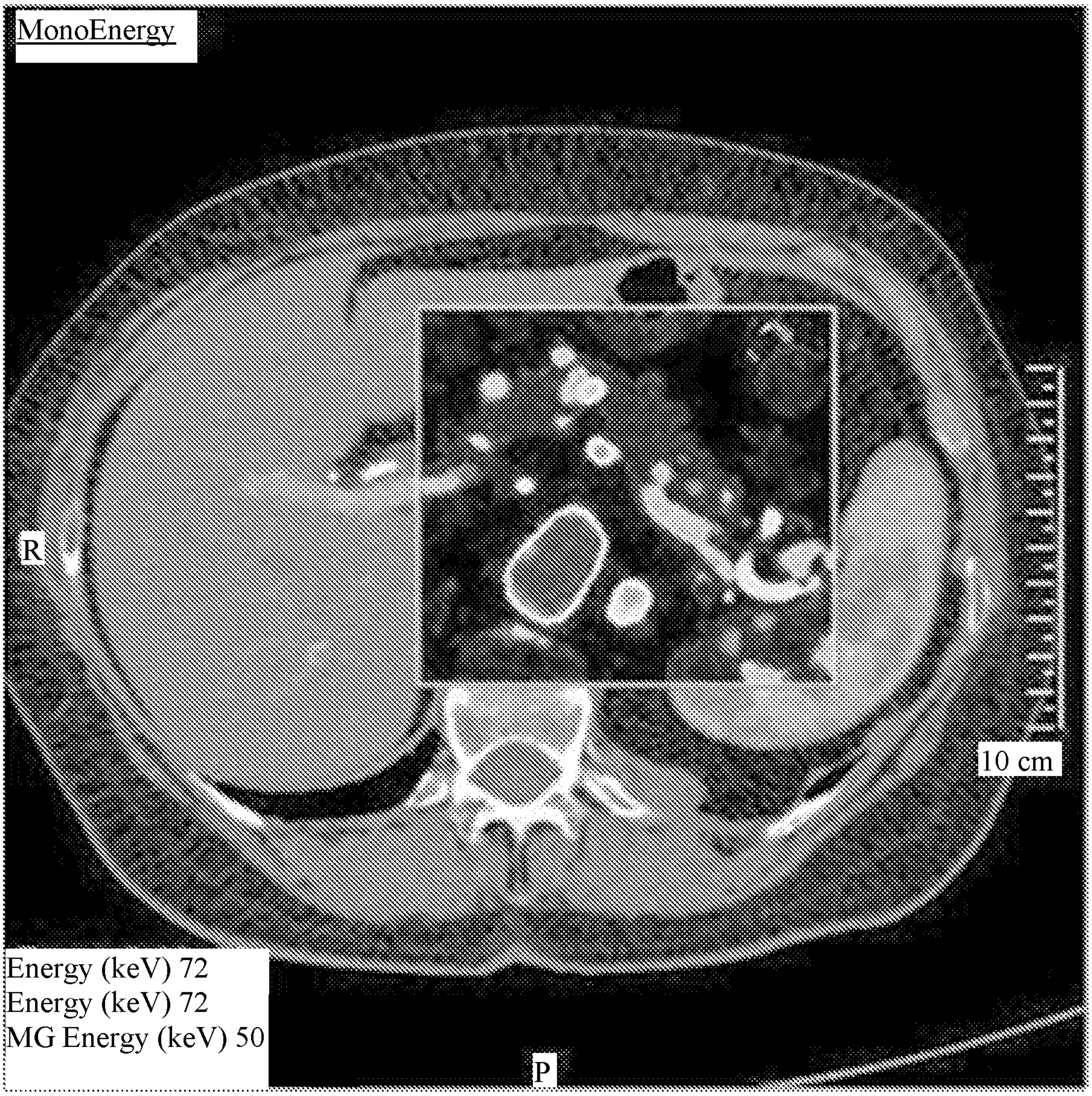

MonoEnergy
R
10 cm
Energy (keV) 72
Energy (keV) 72
MG Energy (keV) 50
P

FIG. 9

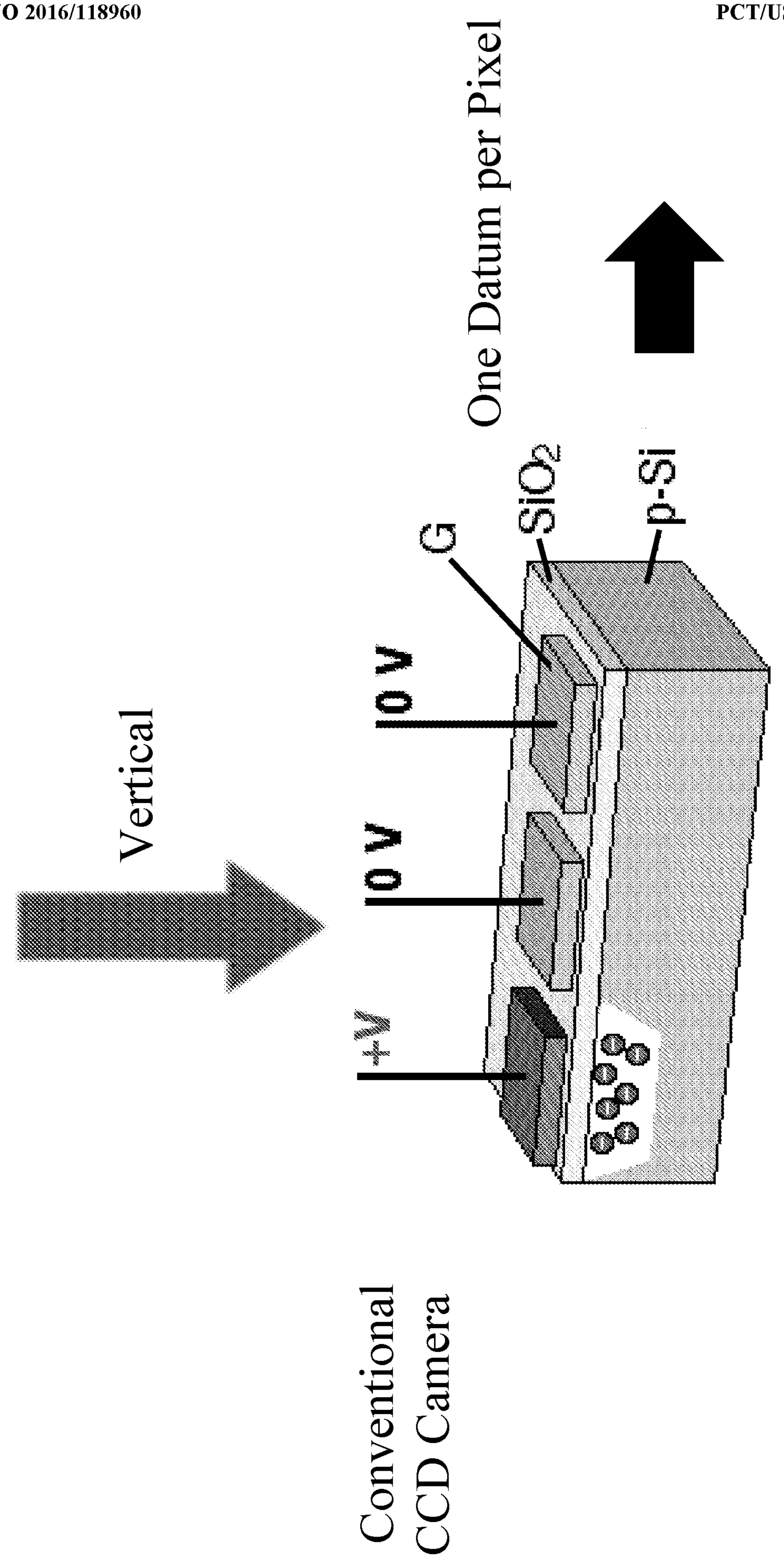

One Datum per Pixel
Vertical
Conventional
CCD Camera
+V
0 V
0 V
G
SiO_2
p-Si

FIG. 10

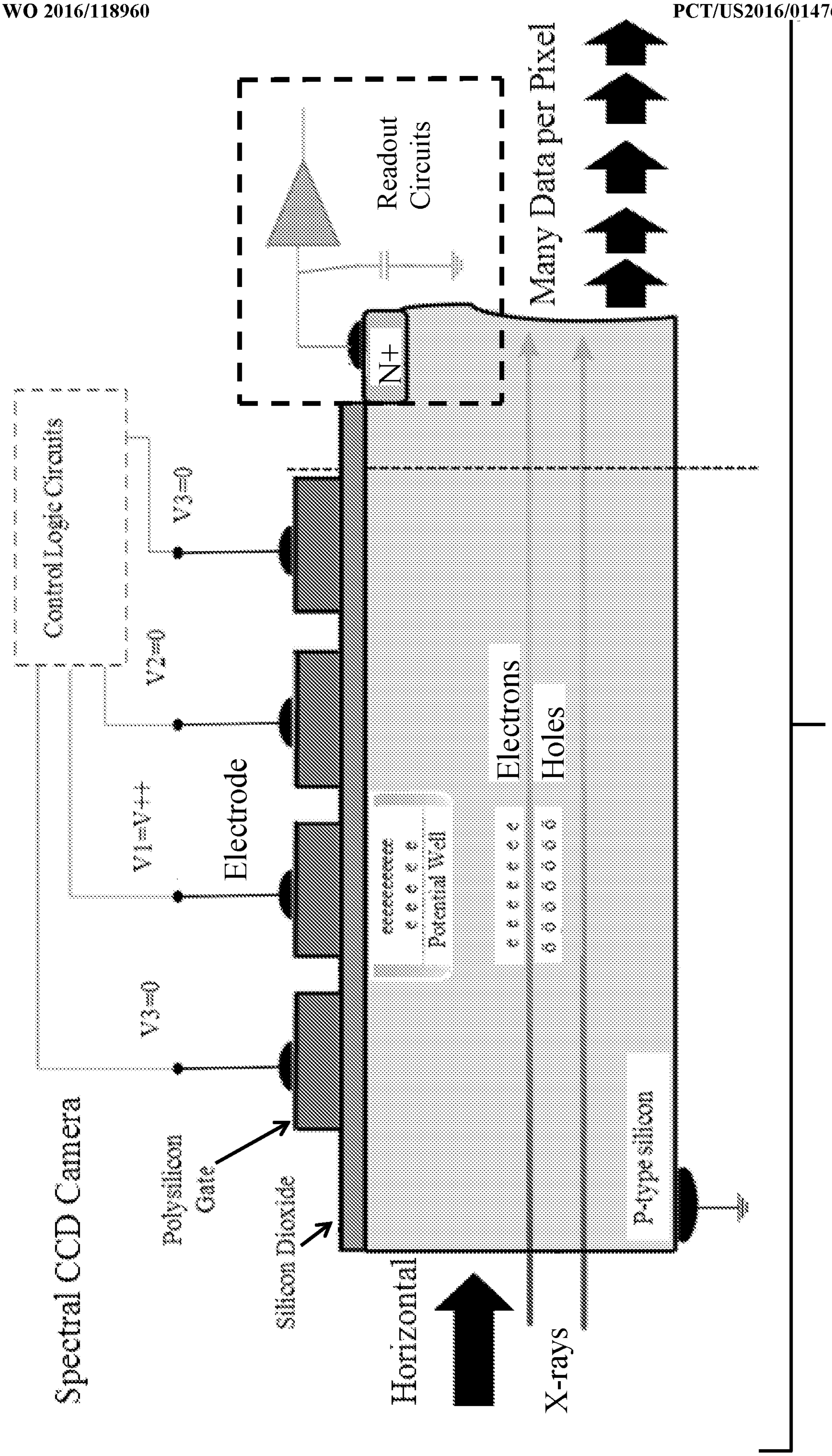

Spectral CCD Camera
Control Logic Circuits
V3=0
V1=V++
V2=0
V3=0
Electrode
Polysilicon Gate
Silicon Dioxide
Potential Well
Electrons
Holes
P-type silicon
N+
Readout Circuits
Many Data per Pixel
Horizontal
X-rays

FIG. 11

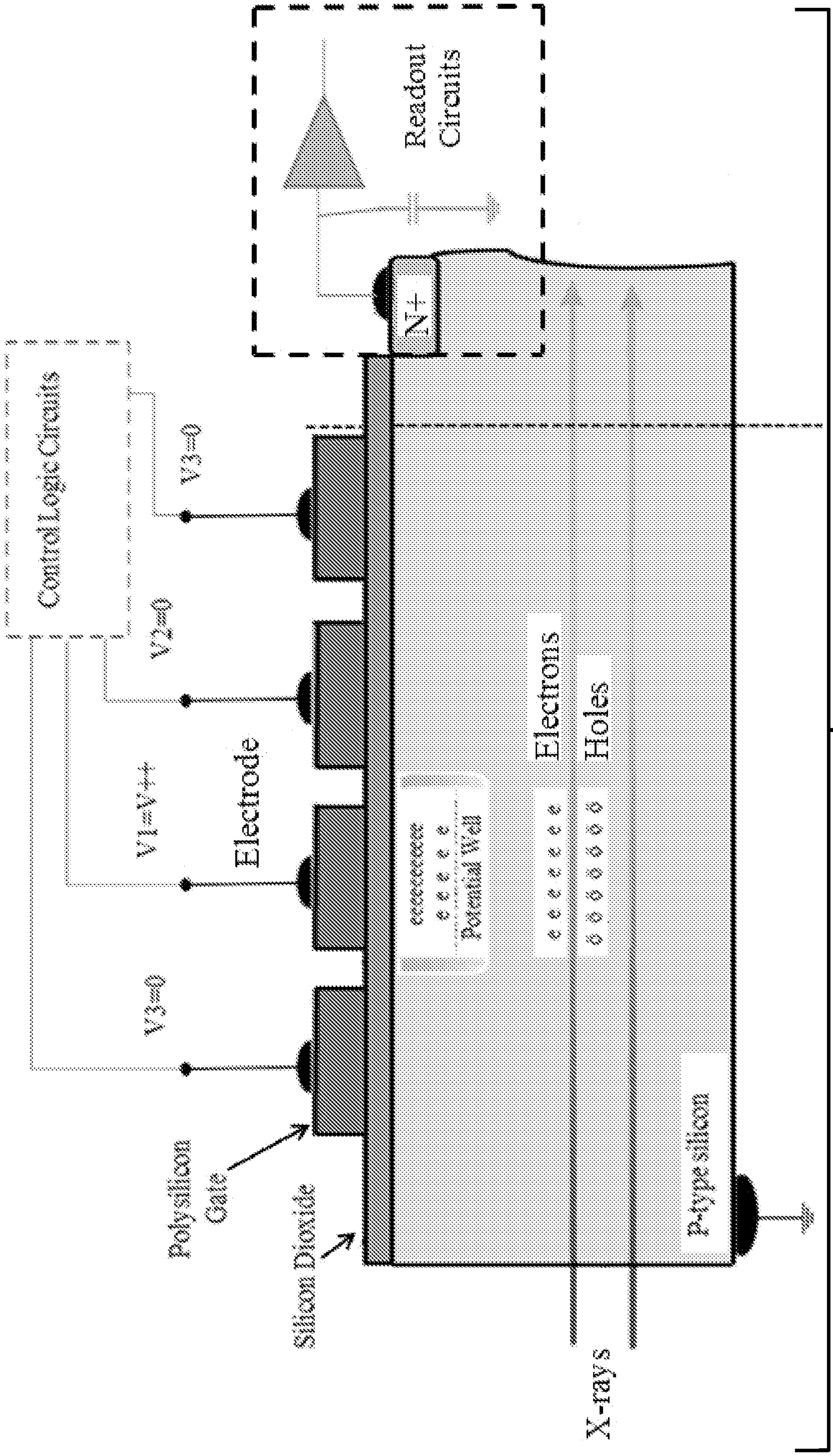
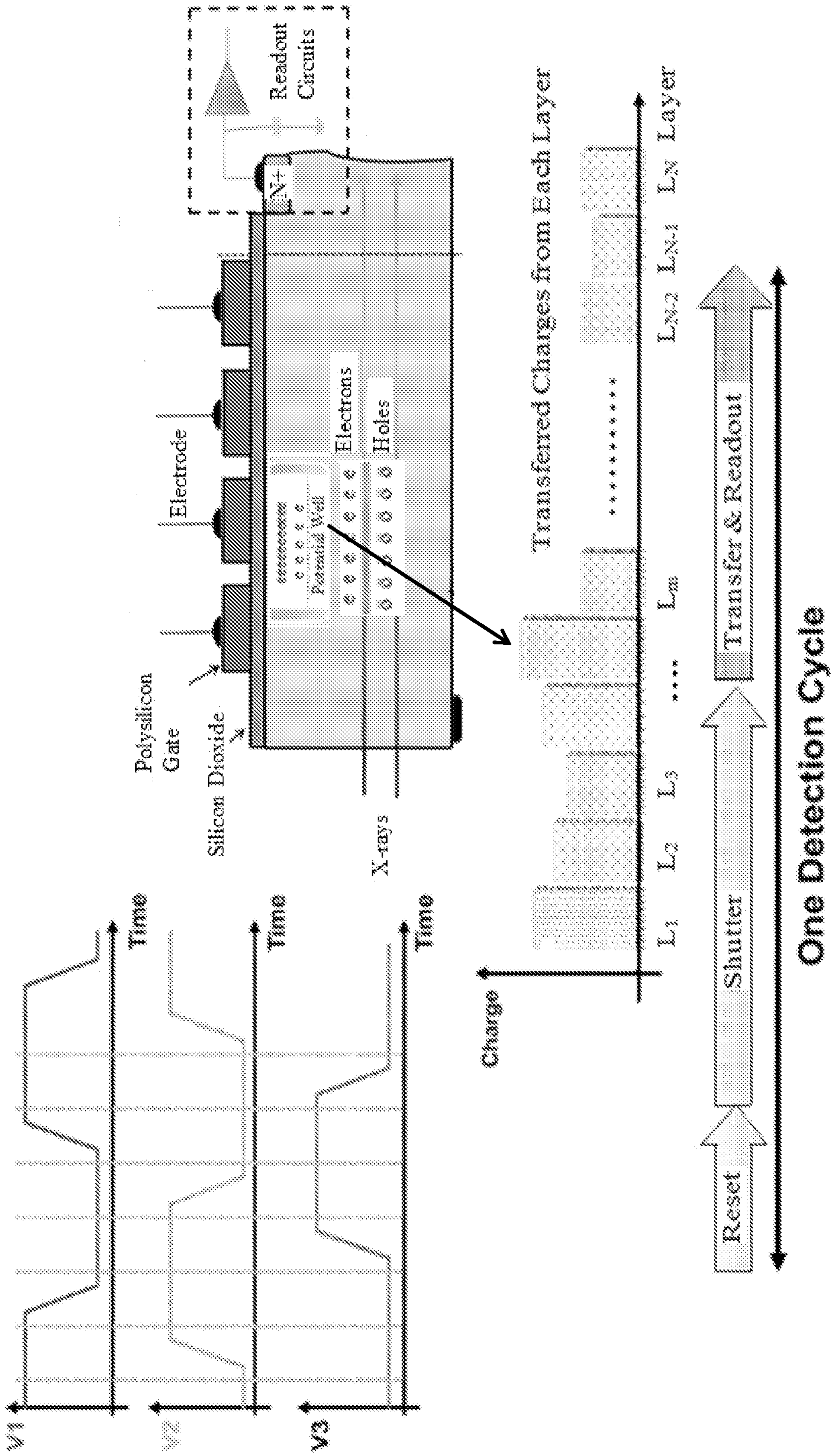

Readout Circuits
N+
Control Logic Circuits
V3=0
V2=0
V1=V++
V3=0
Electrode
Potential Well
Electrons
Holes
Polysilicon Gate
Silicon Dioxide
P-type silicon
X-rays

FIG. 12



V1
V2
V3
Time
Time
Time
Polysilicon Gate
Silicon Dioxide
Electrode
Potential Well
Electrons
Holes
X-rays
N+
Readout Circuits
Transferred Charges from Each Layer
Charge
Layer
L_1
L_2
L_3
L_m
L_{N-2}
L_{N-1}
L_N
Reset
Shutter
Transfer & Readout
One Detection Cycle

FIG. 13

	Medipix3	Detector of subject invention
Dynamic Range	14 bit	80 DB / 14 bit
Acquisition Time	1 ms	40 μs – 1 ms
Frame Rate	500 fps	1k fps
Pixel Size	110 x 110 μm^2 55 x55 μm^2	650 x650 μm^2
Fill Factor	>90%	85.2%
Material & Thickness	2 mm CdTe 2 mm CZT 300 μm Si 300-900 μm GaAs	33mm Si
Energy Range	10-50 keV	10-50 keV
Working Mode	Photon Counting	PE & Charge Coupling

FIG. 14

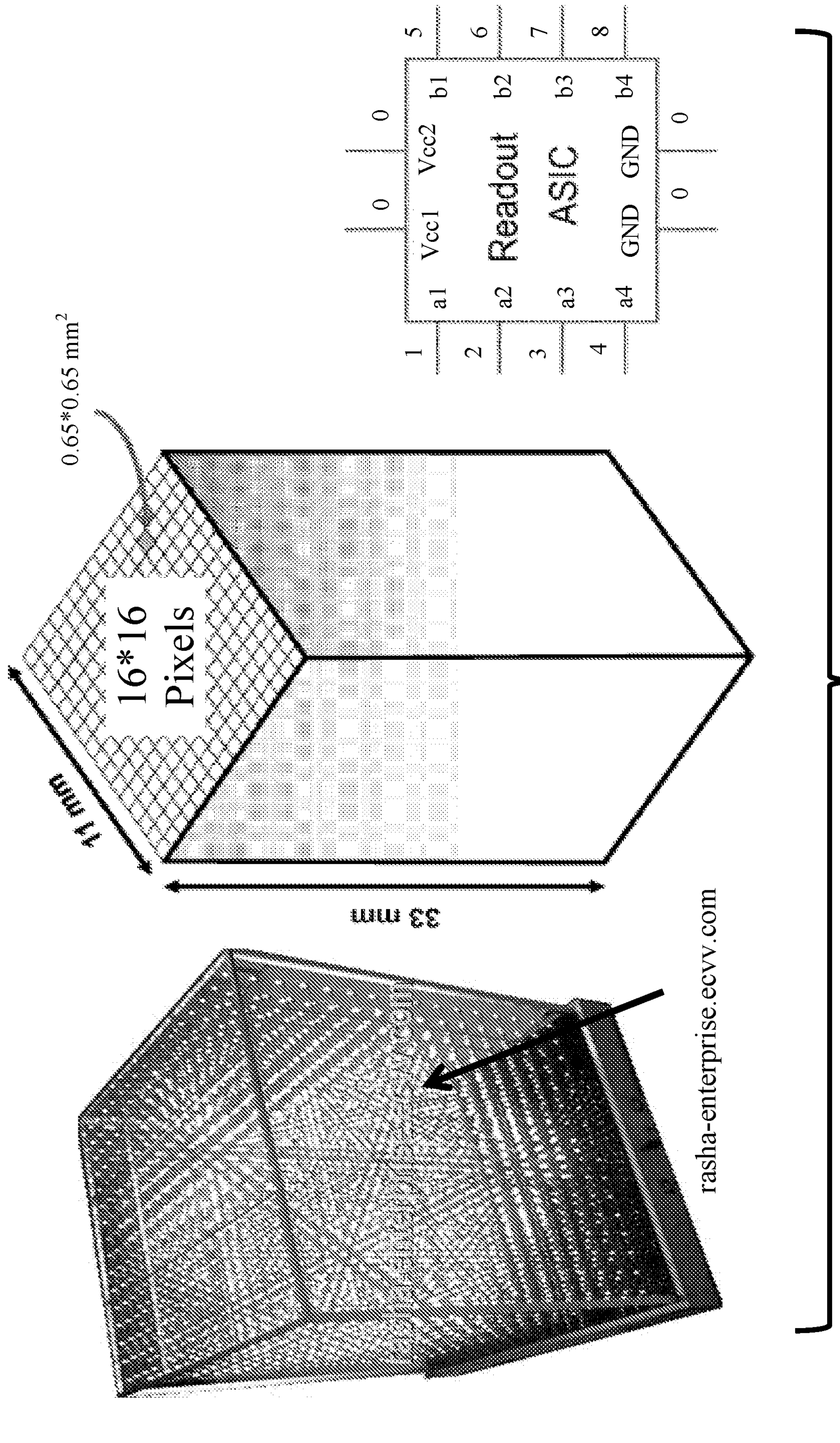

rasha-enterprise.ecvv.com
33 mm
11 mm
16*16
Pixels
0.65*0.65 mm²
Readout
ASIC
Vcc1
Vcc2
GND
GND
a1
a2
a3
a4
b1
b2
b3
b4
1
2
3
4
5
6
7
8
0
0
0
0

FIG. 15

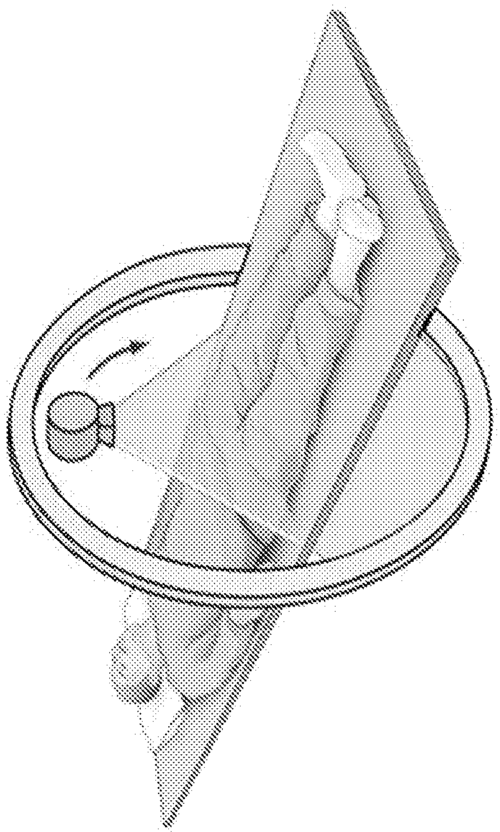


FIG. 16A

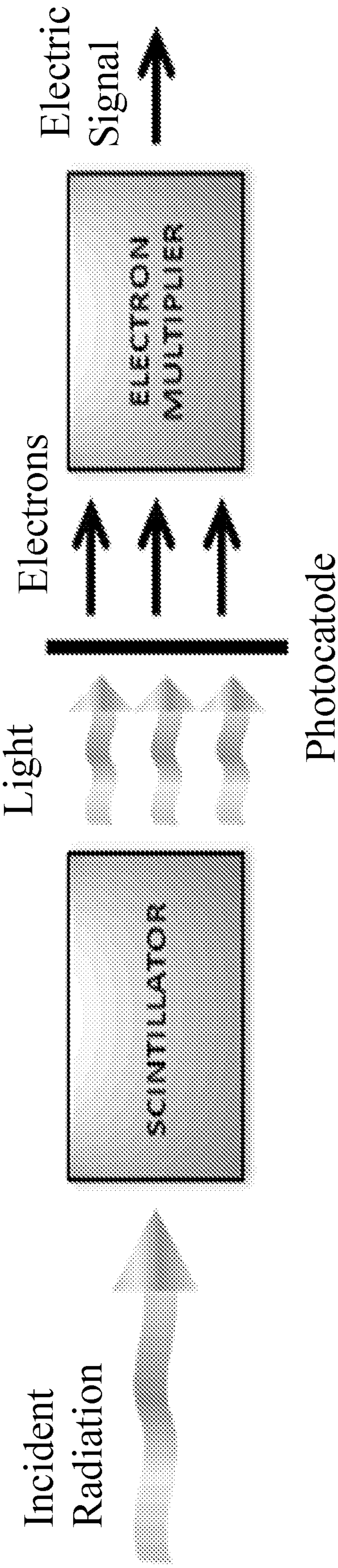


FIG. 16B

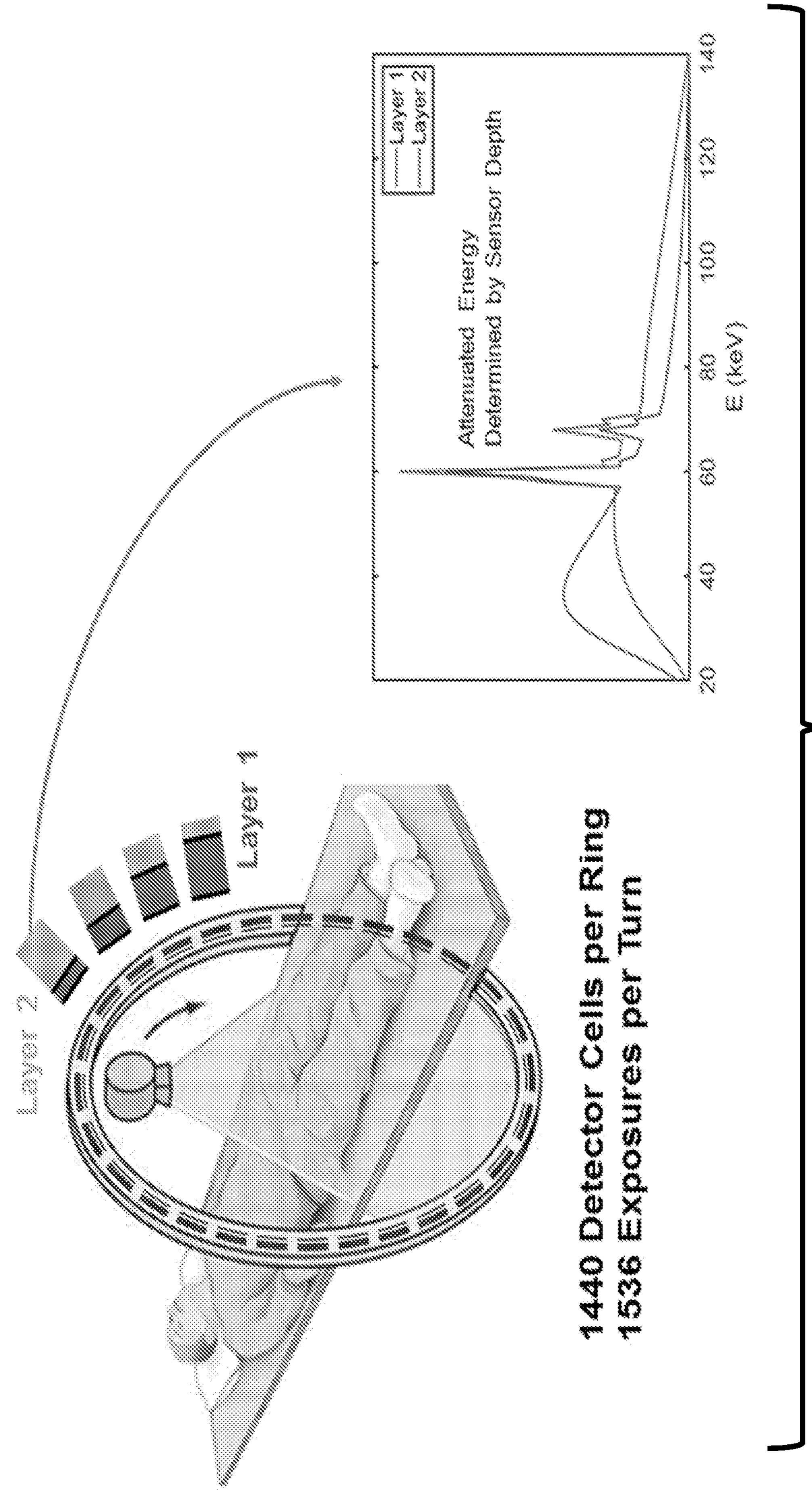

Layer 2
Layer 1
1440 Detector Cells per Ring
1536 Exposures per Turn
Attenuated Energy
Determined by Sensor Depth
Layer 1
Layer 2
20
40
60
80
100
120
140
E (keV)

FIG. 17

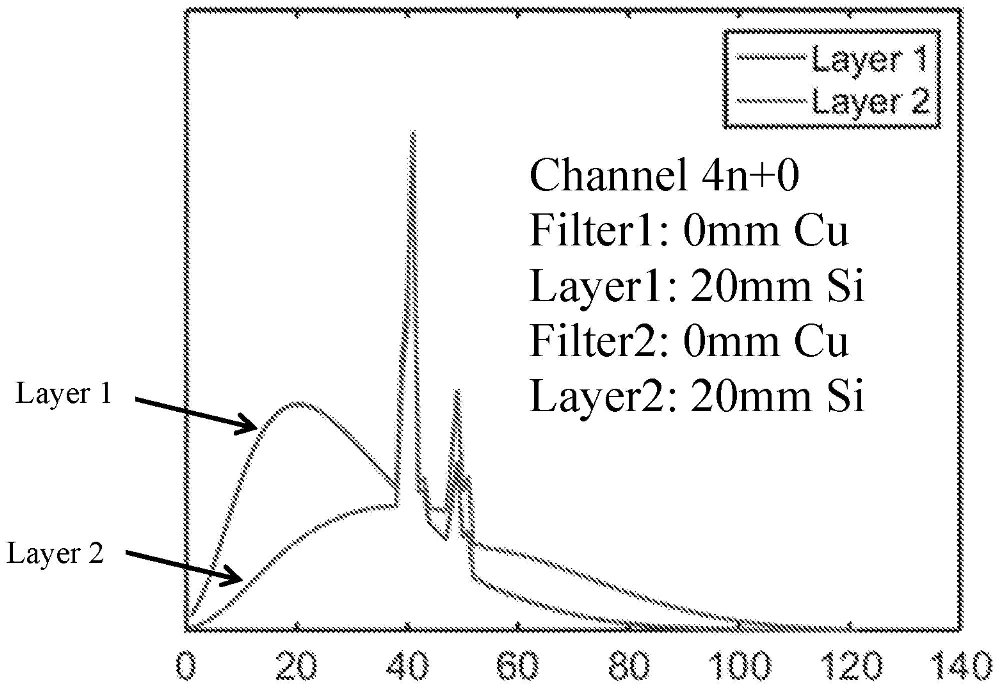


FIG. 18A

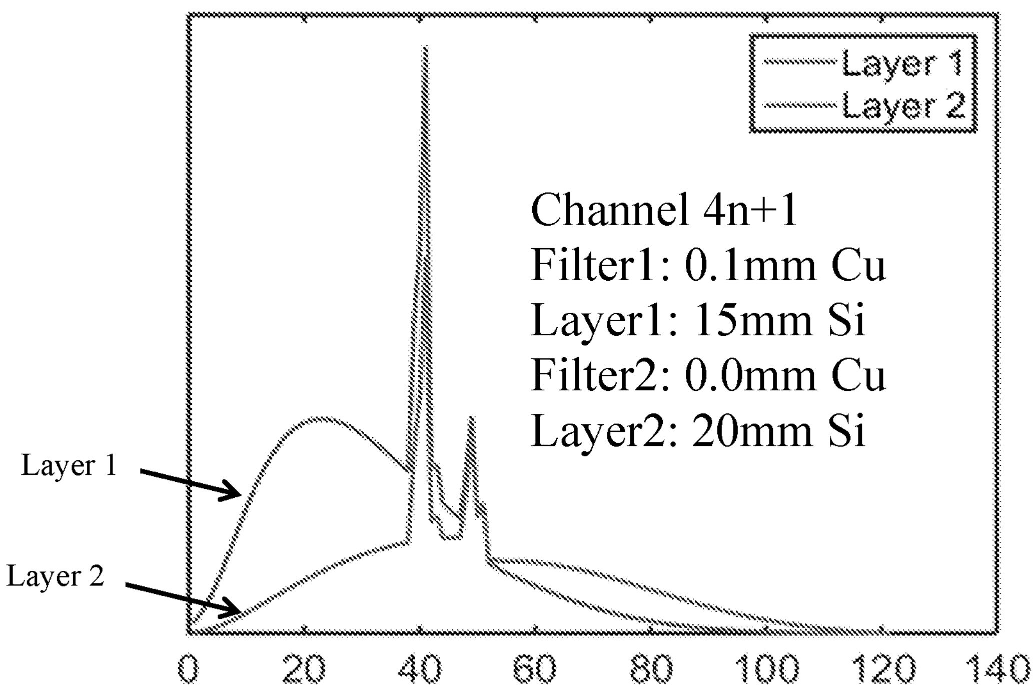


FIG. 18B

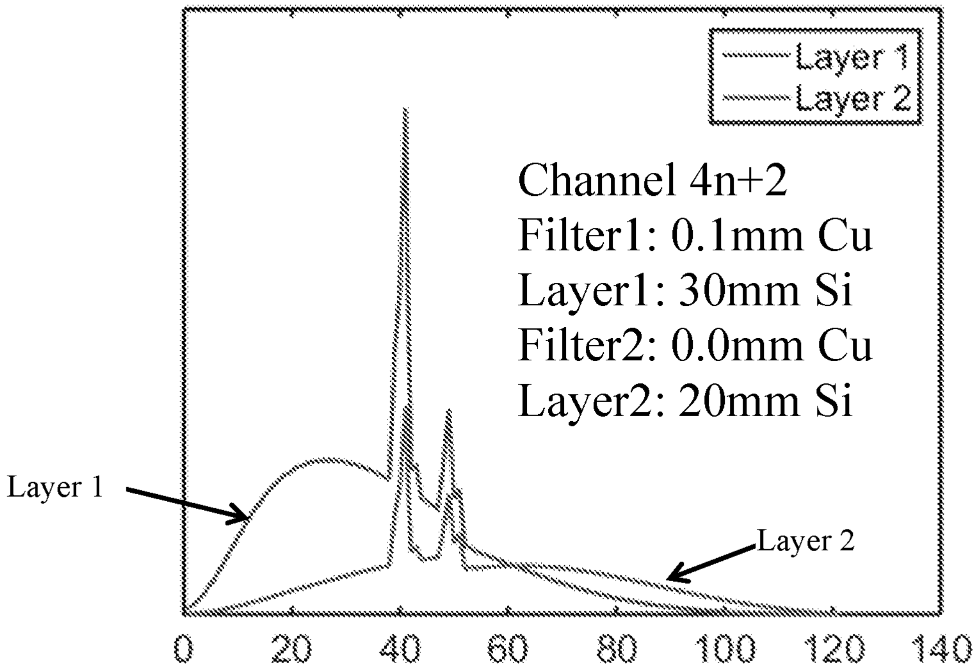


FIG. 18C

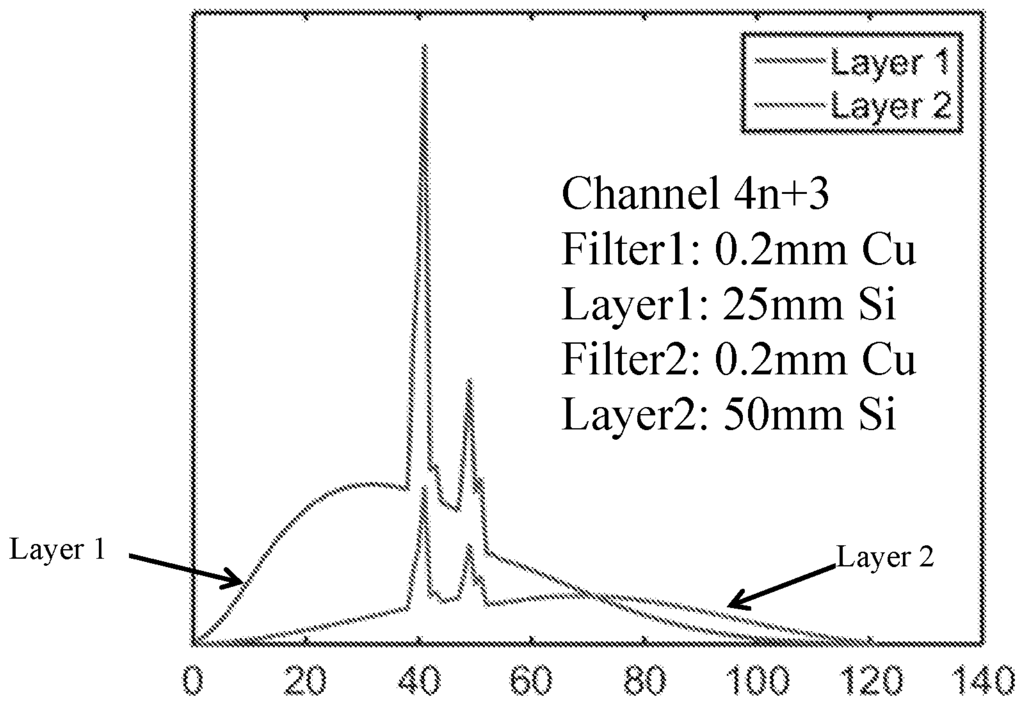


FIG. 18D

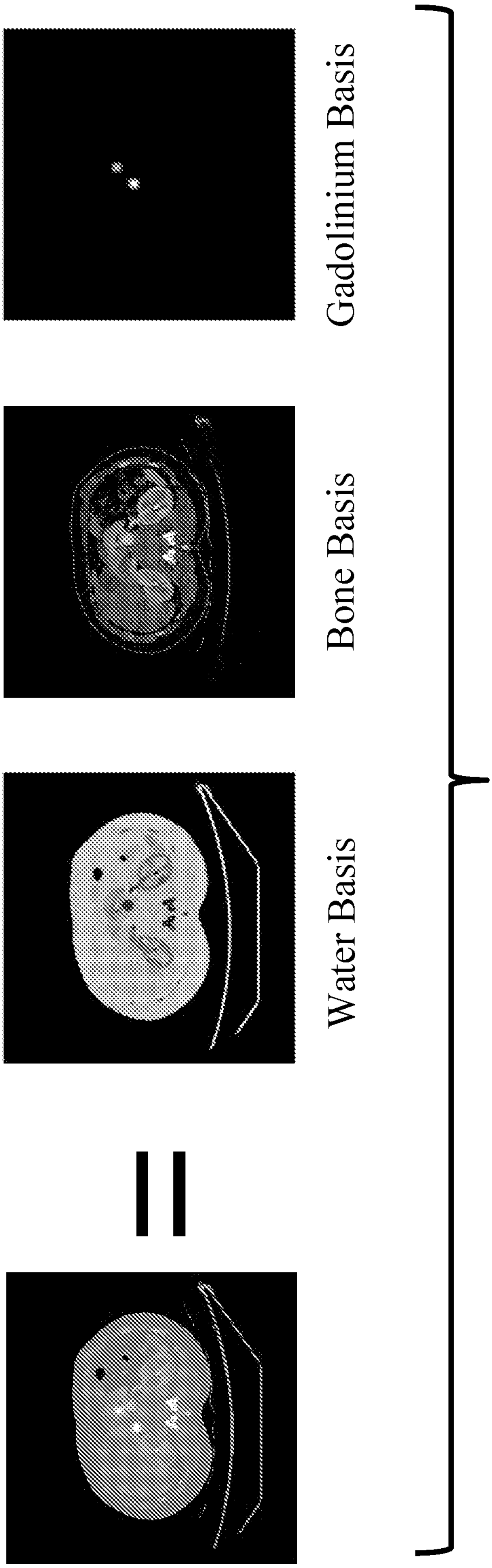
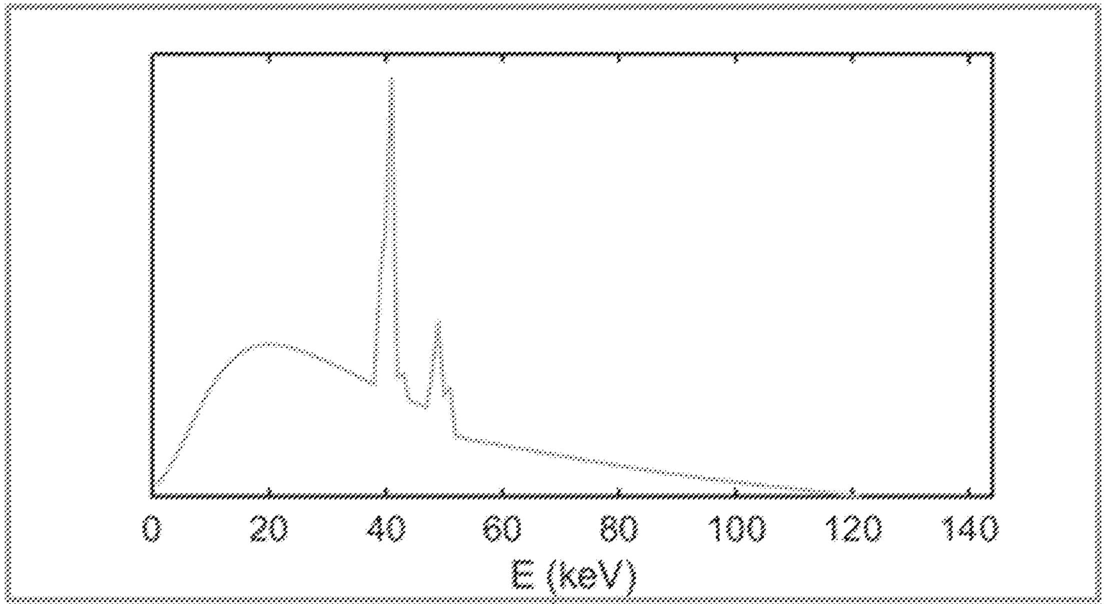
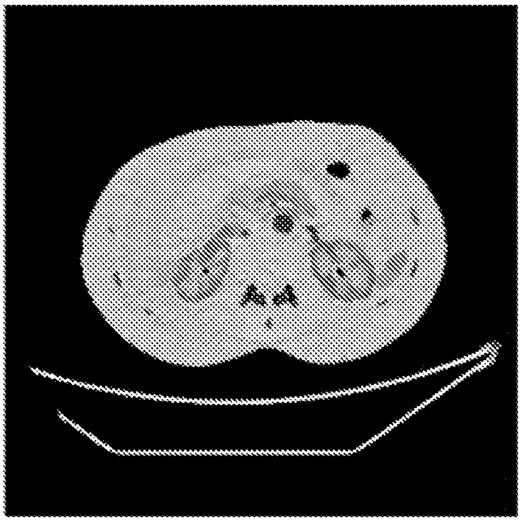


FIG. 19



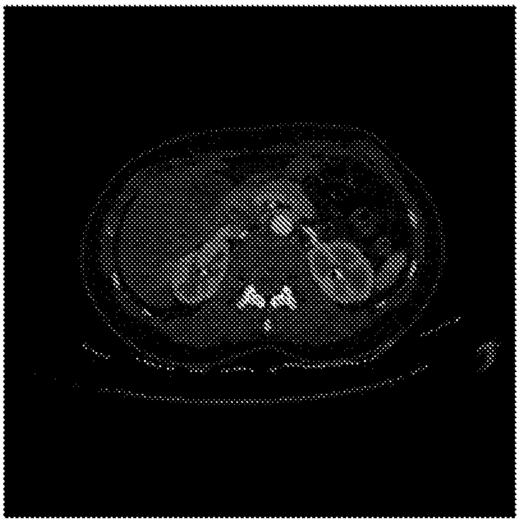
$$I = \sum_{k=1}^{K} S_k \cdot \exp\left(-\sum_{j=1}^{J} l_j \left(a_1 \cdot \mu_{water} + a_2 \cdot \mu_{bone} + a_3 \cdot \mu_{Gd} \right) \right)$$

FIG. 20



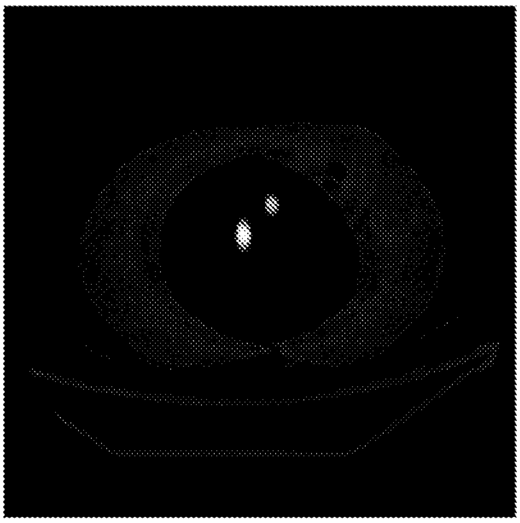
Water Basis

FIG. 21A



Bone Basis

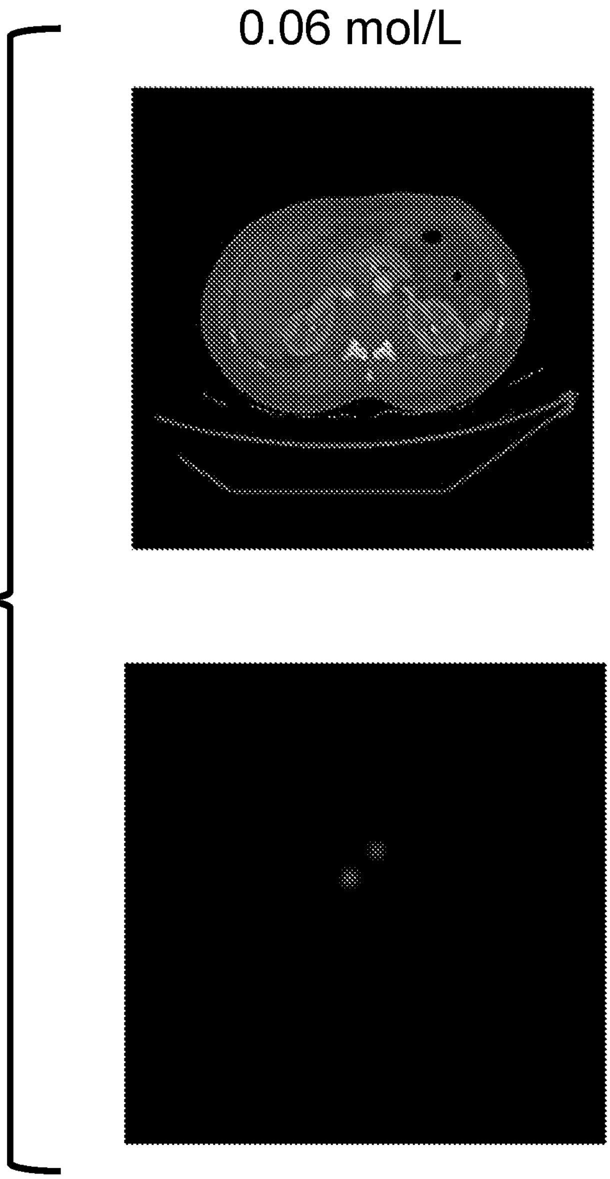
FIG. 21B



Gadolinium Basis

FIG. 21C

FIG. 22A



0.09 mol/L

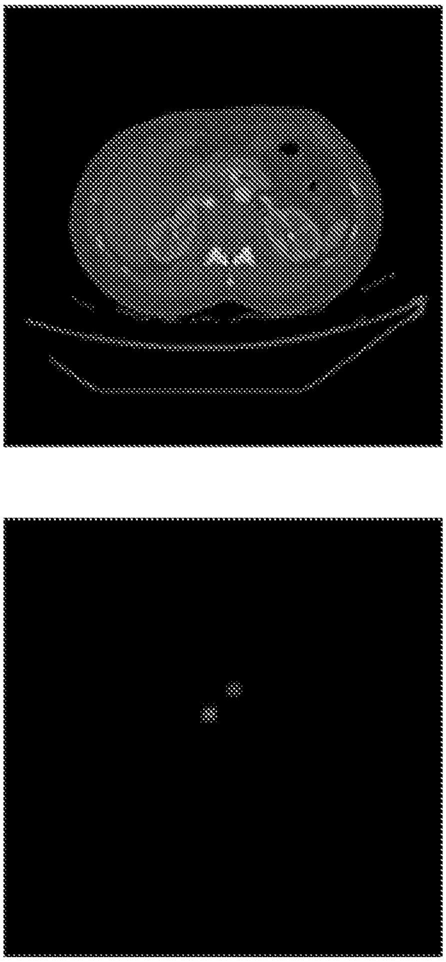


FIG. 22B

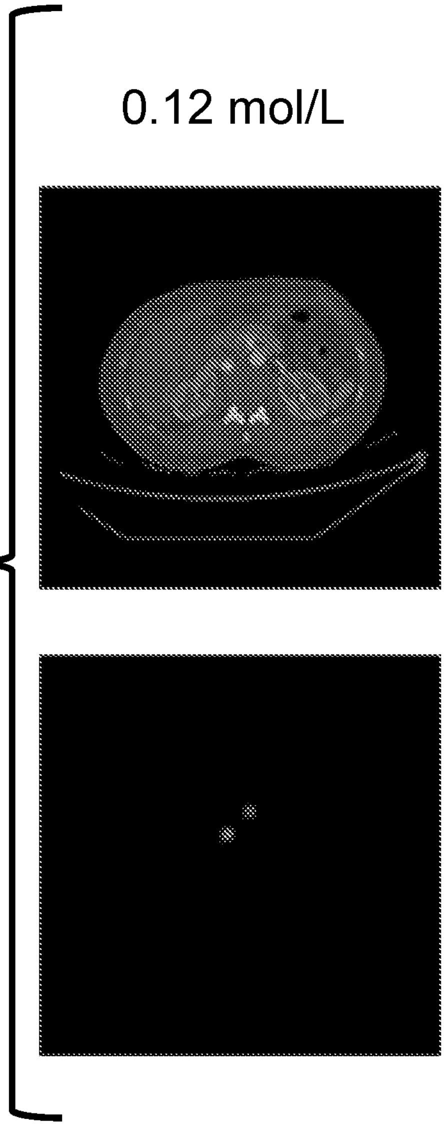
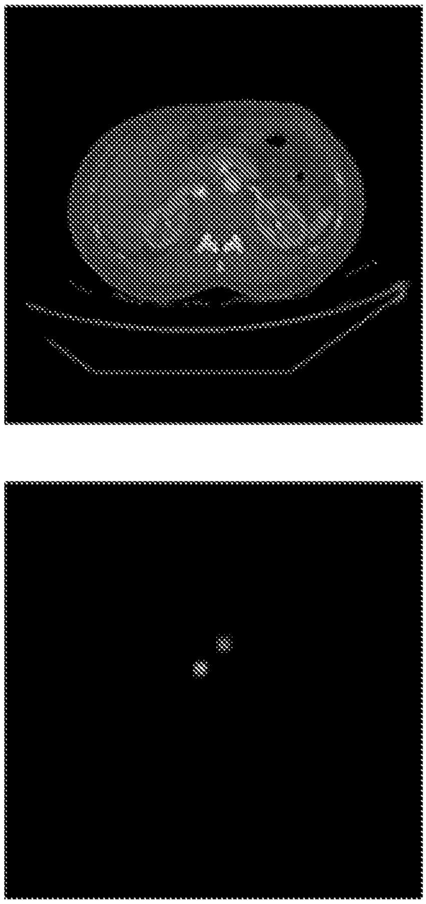


FIG. 22C

0.15 mol/L

FIG. 22D



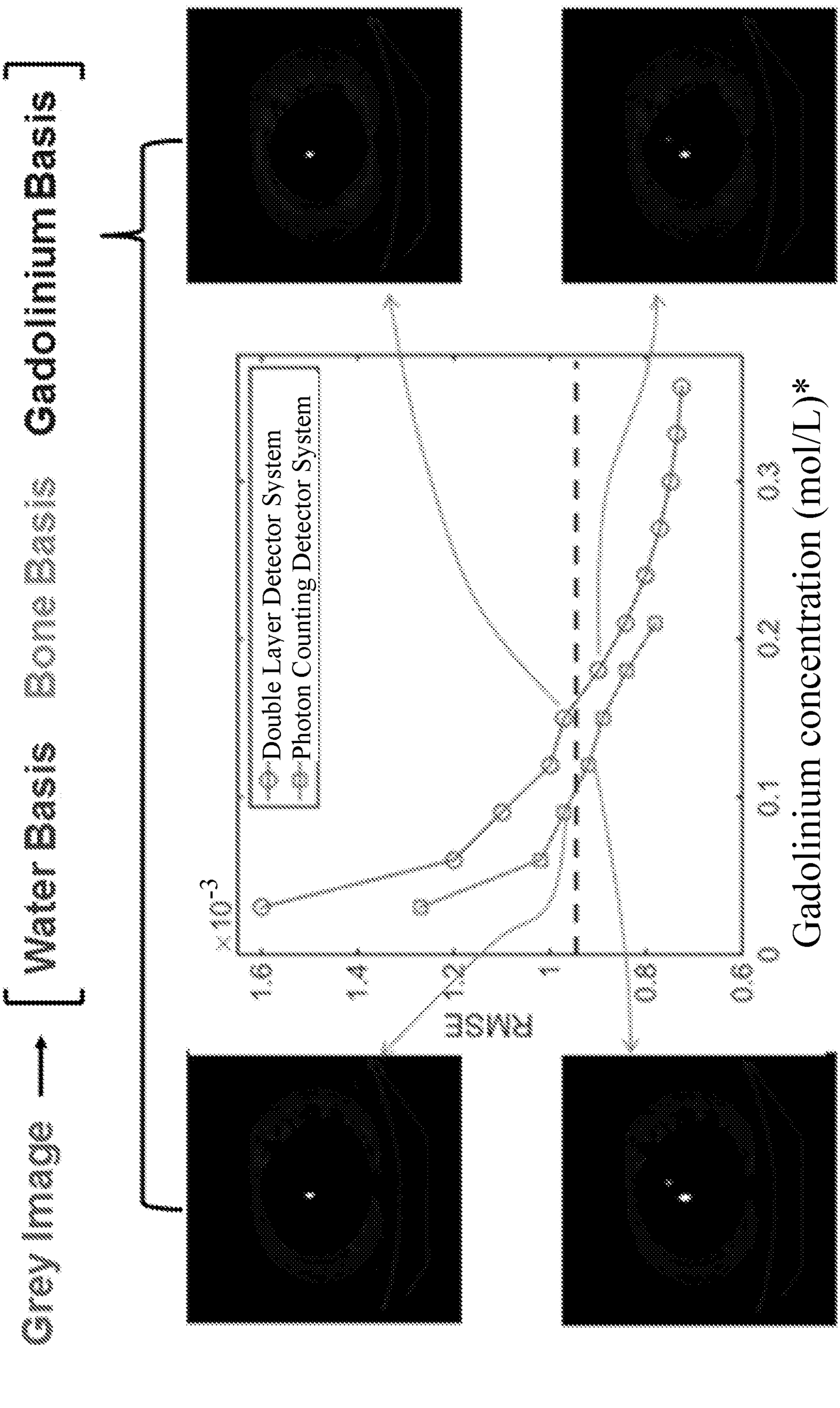

Grey Image → [Water Basis Bone Basis Gadolinium Basis]
RMSE
$\times 10^{-3}$
1.6
1.4
1.2
1
0.8
0.6
0
0.1
0.2
0.3
Double Layer Detector System
Photon Counting Detector System
Gadolinium concentration (mol/L)*

FIG. 23

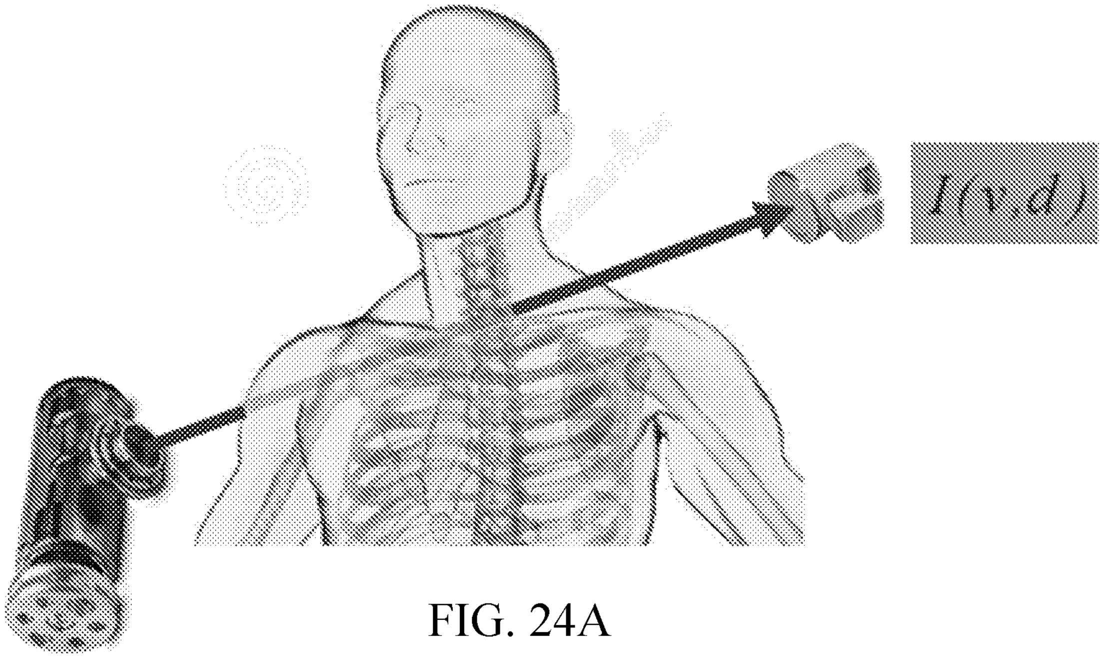


FIG. 24A

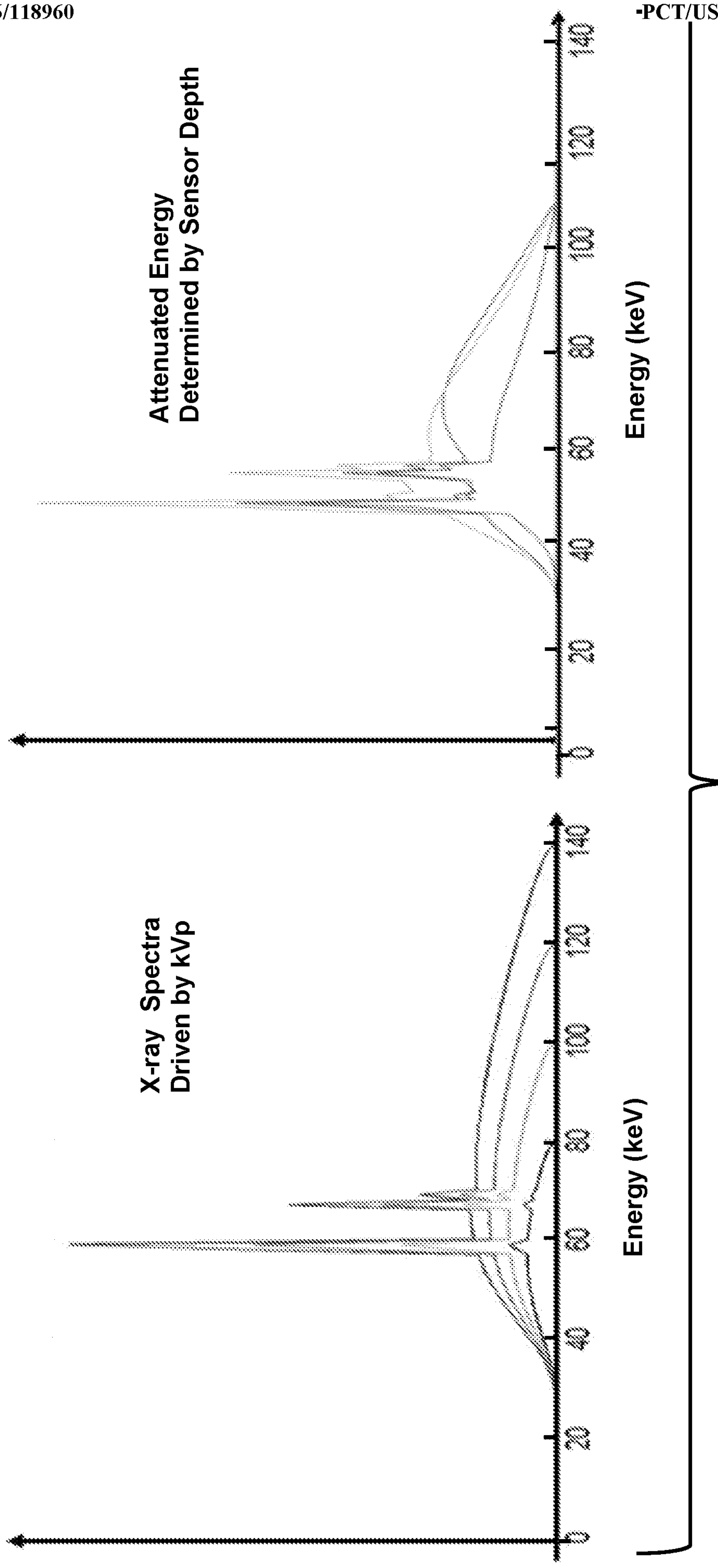

Attenuated Energy
Determined by Sensor Depth
Energy (keV)
0
20
40
60
80
100
120
140
X-ray Spectra
Driven by kVp
Energy (keV)
0
20
40
60
80
100
120
140

FIG. 24B

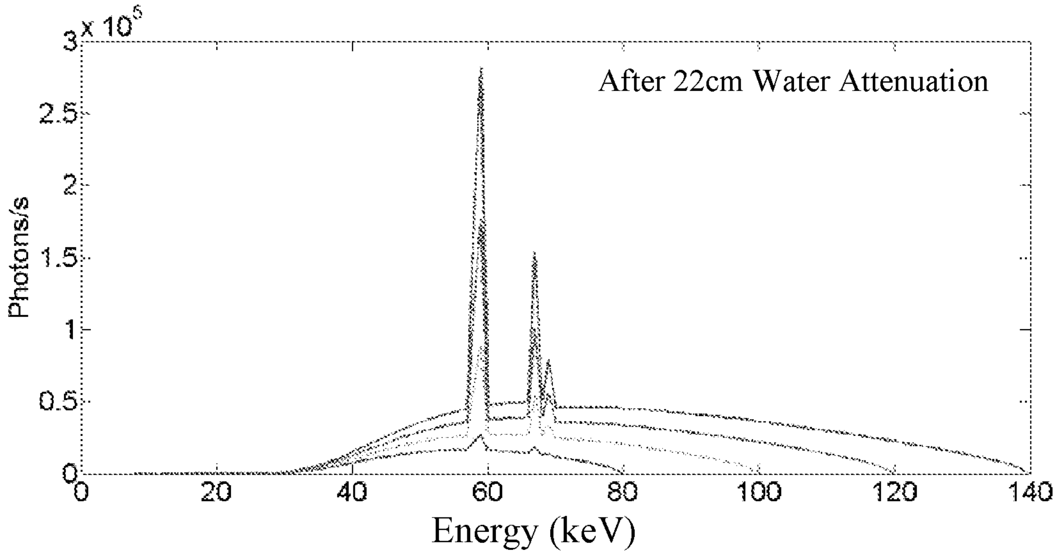


FIG. 25

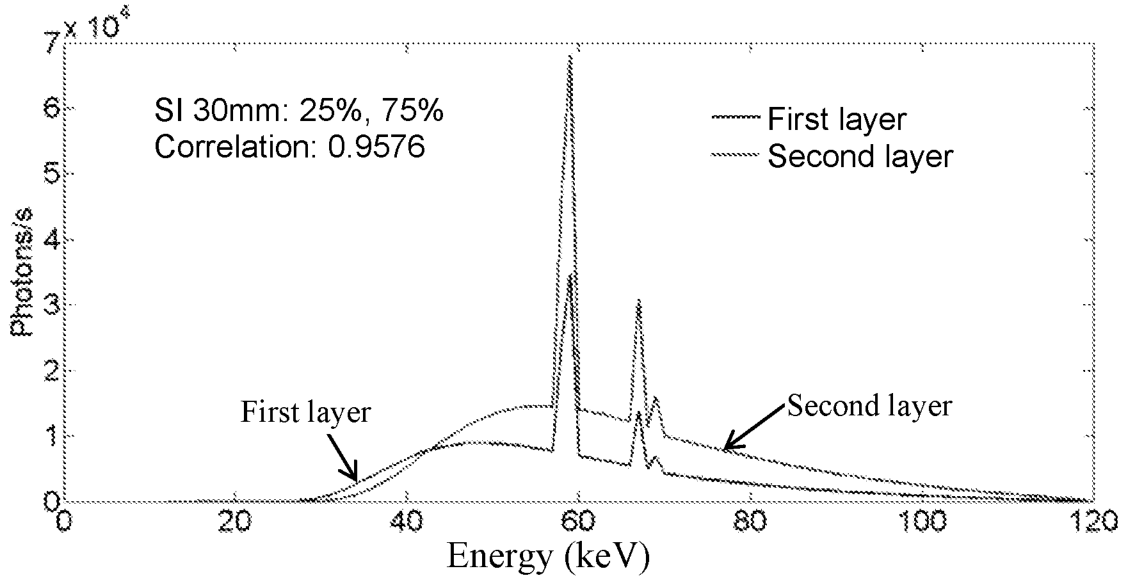


FIG. 26A

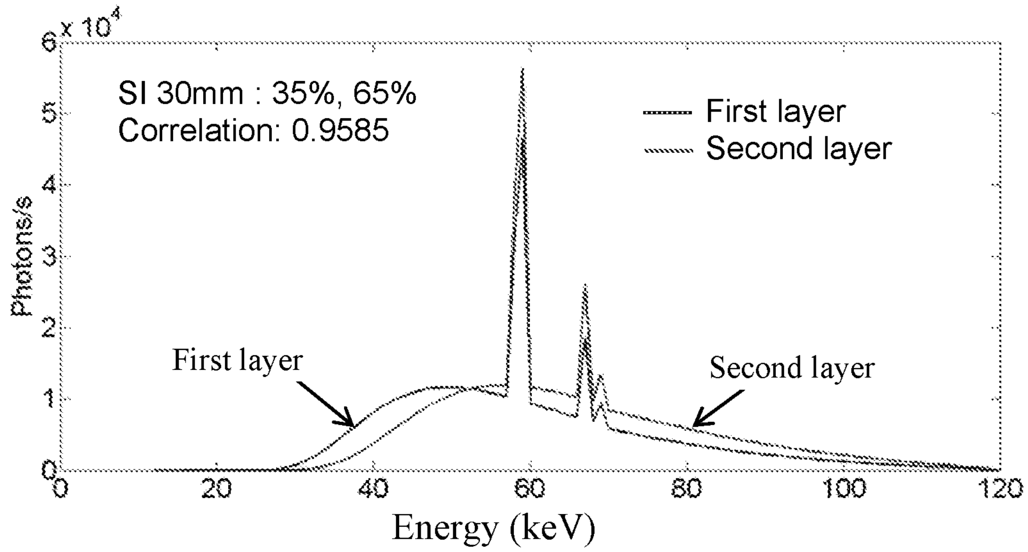


FIG. 26B

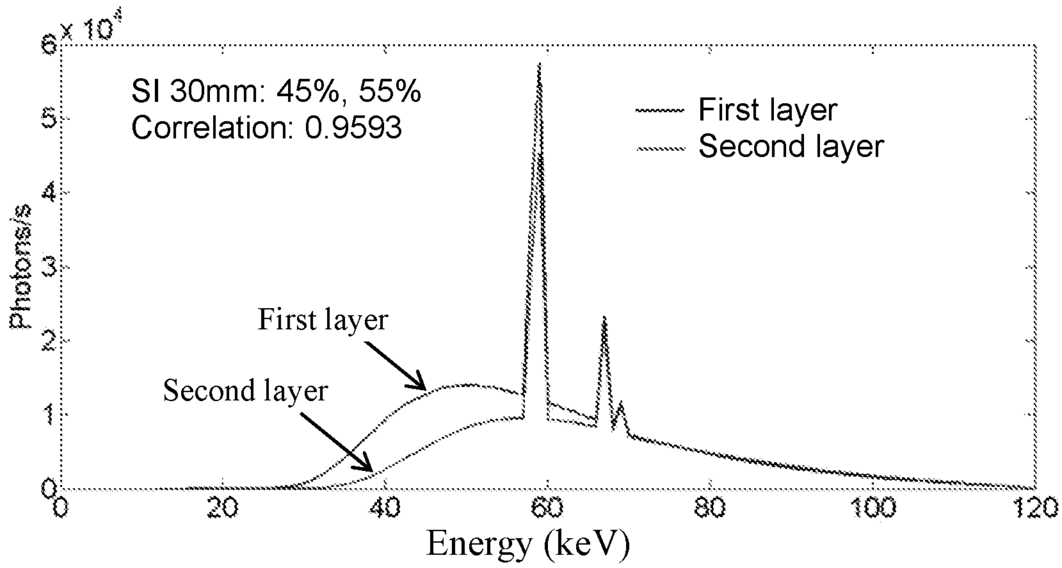


FIG. 26C

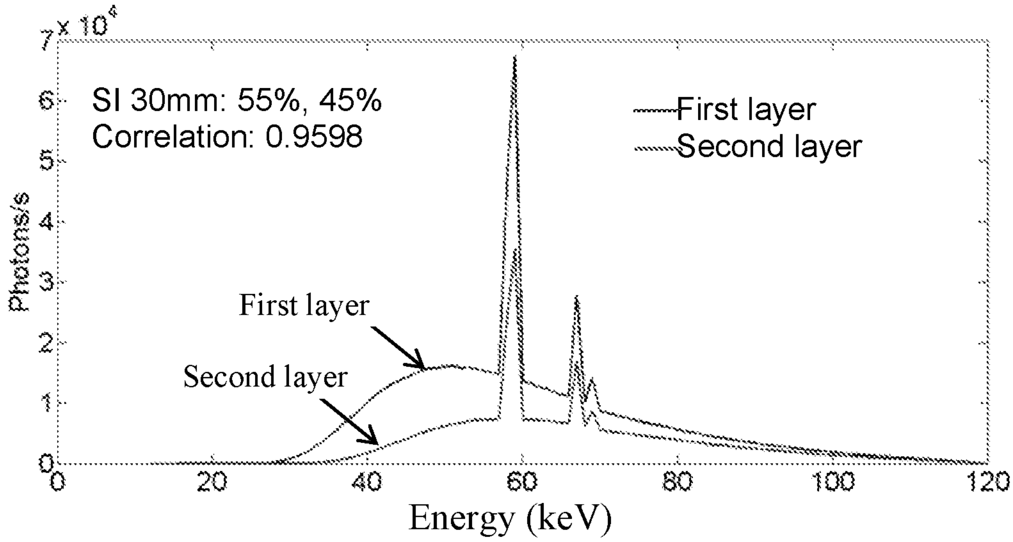


FIG. 26D

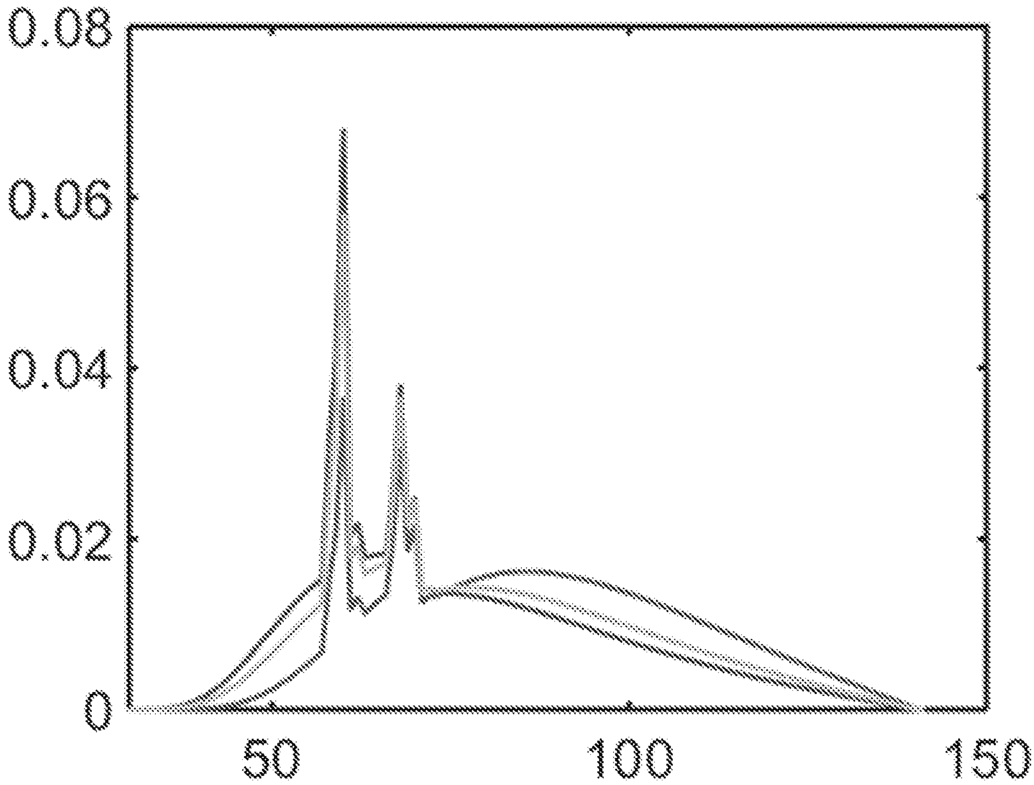


FIG. 27A

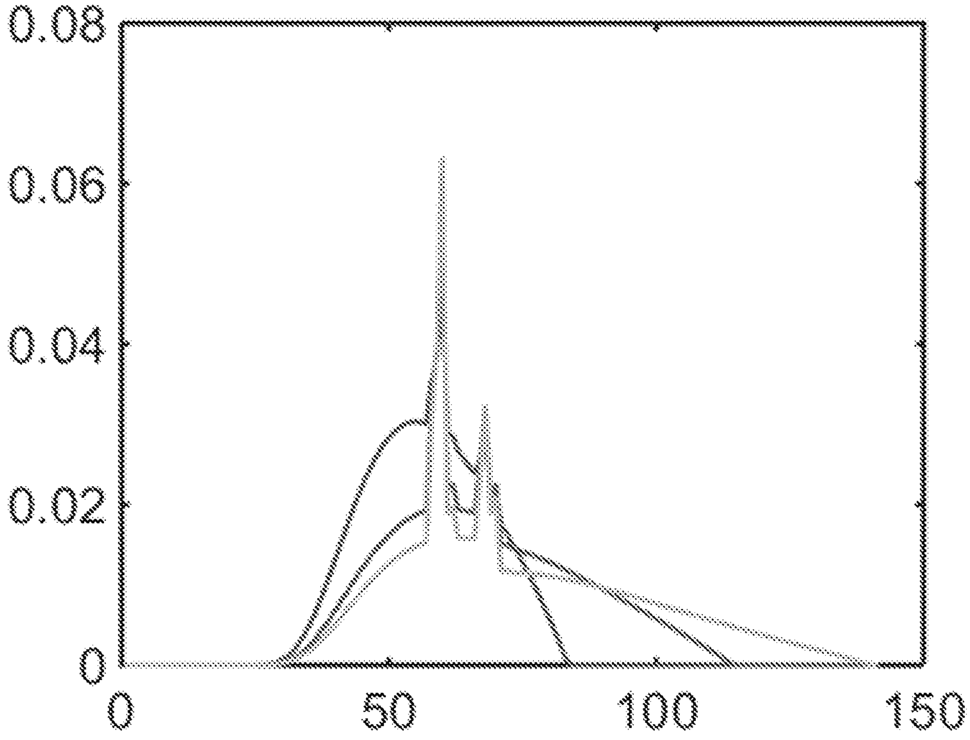


FIG. 27B

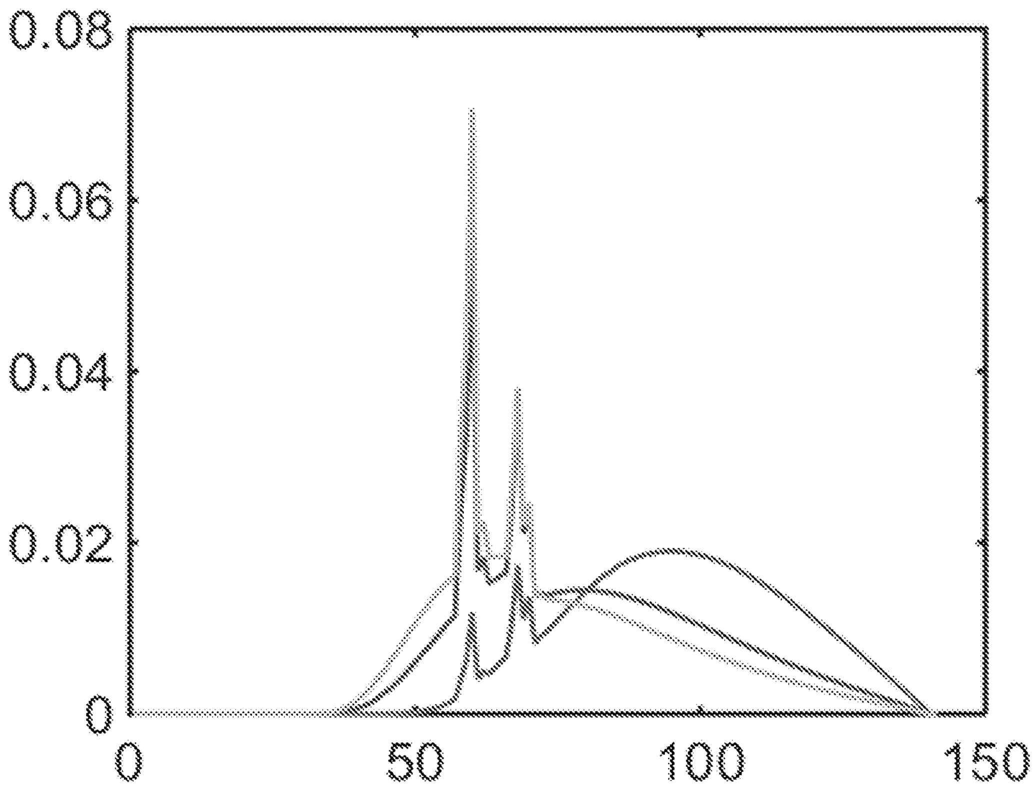


FIG. 27C

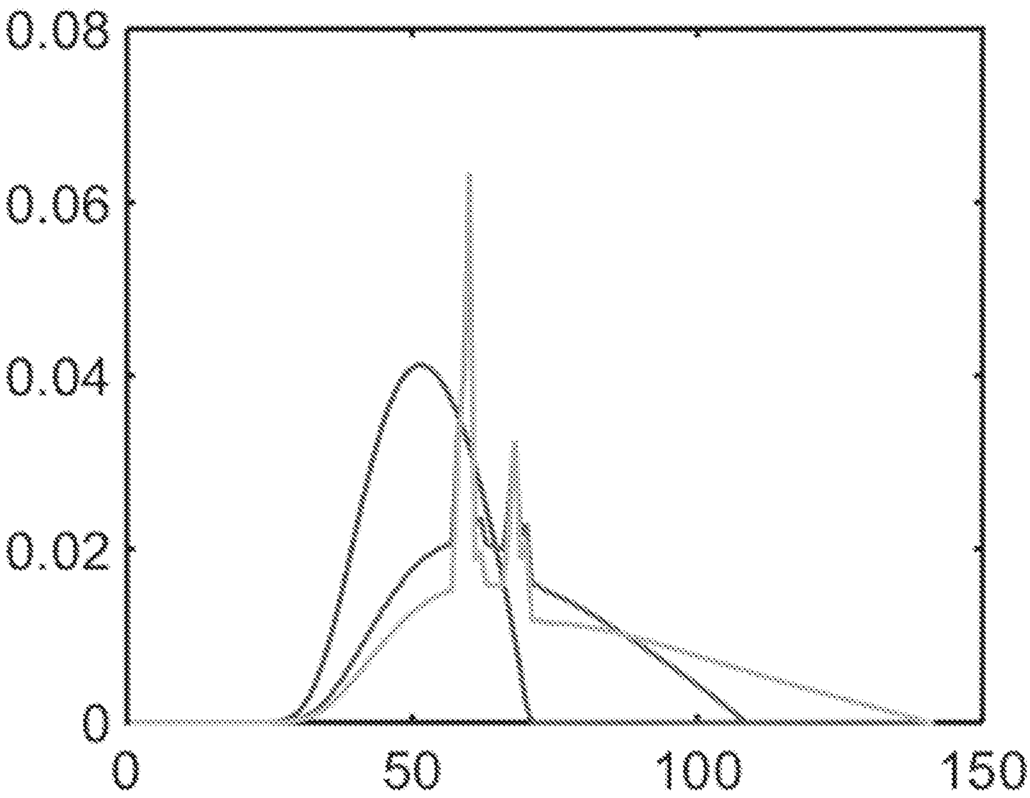


FIG. 27D

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US2016/014769

A. CLASSIFICATION OF SUBJECT MATTER

G01T 1/36(2006.01)i, G01T 1/29(2006.01)i, G01T 1/15(2006.01)i

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
G01T 1/36; A61B 6/14; G01T 1/16; G21K 1/12; A61B 6/03; G06K 9/00; G01T 1/24; G01T 1/29; G01T 1/15

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched
Korean utility models and applications for utility models
Japanese utility models and applications for utility models

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)
eKOMPASS(KIPO internal) & Keywords: electrode, compute, scanner, x-ray, source, detector, control and tomography

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 2005-0226361 A1 (ZHOU, OTTO Z. et al.) 13 October 2005 See paragraphs [0014]-[0049]; claims 1, 25; and figure 3.	1-3,32-35
A	WO 2014-028930 A1 (THE UNIVERSITY OF NORTH CAROLINA AT CHAPEL HILL) 20 February 2014 See claims 1-21.	1-3,32-35
A	US 2010-0239064 A1 (ZHOU, OTTO Z. et al.) 23 September 2010 See paragraphs [0025]-[0034] and figures 1-4.	1-3,32-35
A	US 2012-0205549 A1 (SIMON, MATTHIAS et al.) 16 August 2012 See paragraphs [0047]-[0055] and figures 1-5.	1-3,32-35
A	US 2009-0321643 A1 (RUTTEN, WALTER et al.) 31 December 2009 See paragraphs [0071]-[0093] and figures 1-4.	1-3,32-35

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

* Special categories of cited documents:
"A" document defining the general state of the art which is not considered to be of particular relevance
"E" earlier application or patent but published on or after the international filing date
"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed
"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents,such combination being obvious to a person skilled in the art
"&" document member of the same patent family

Date of the actual completion of the international search
25 April 2016 (25.04.2016)

Date of mailing of the international search report
25 April 2016 (25.04.2016)

Name and mailing address of the ISA/KR
International Application Division
Korean Intellectual Property Office
189 Cheongsa-ro, Seo-gu, Daejeon, 35208, Republic of Korea
Facsimile No. +82-42-481-8578

Authorized officer
KIM, Jin Ho
Telephone No. +82-42-481-8699

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US2016/014769

Box No. II Observations where certain claims were found unsearchable (Continuation of item 2 of first sheet)

This international search report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

1. ☐ Claims Nos.:
because they relate to subject matter not required to be searched by this Authority, namely:

2. ☒ Claims Nos.: 15, 18, 23, 41, 54, 57, 62, 71-72
because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out, specifically:
The claims 15, 18, 23, 41, 54, 57, 71-72 are regarded unclear since they refer to multiple dependent claims which do not comply with PCT Rule 6.4(a).
Claim 62 is referring to claim 101 which does not exist in this application.

3. ☒ Claims Nos.: 4-14, 16-17, 19-22, 24-31, 36-40, 42-53, 55-56, 58-61, 63-70, 73
because they are dependent claims and are not drafted in accordance with the second and third sentences of Rule 6.4(a).

Box No. III Observations where unity of invention is lacking (Continuation of item 3 of first sheet)

This International Searching Authority found multiple inventions in this international application, as follows:

1. ☐ As all required addtional search fees were timely paid by the applicant, this international search report covers all searchable claims.

2. ☐ As all searchable claims could be searched without effort justifying an additional fees, this Authority did not invite payment of any additional fees.

3. ☐ As only some of the required additional search fees were timely paid by the applicant, this international search report covers only those claims for which fees were paid, specifically claims Nos.:

4. ☐ No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claims Nos.:

Remark on Protest

☐ The additional search fees were accompanied by the applicant's protest and, where applicable, the payment of a protest fee.

☐ The additional search fees were accompanied by the applicant's protest but the applicable protest fee was not paid within the time limit specified in the invitation.

☐ No protest accompanied the payment of additional search fees.

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US 2005-0226361 A1	13/10/2005	AU 2002-12979 A1	22/04/2002
		AU 2003-202930 A1	02/09/2003
		CA 2424826 A1	18/04/2002
		CA 2474053 A1	31/07/2003
		CN 100459019 C	04/02/2009
		CN 101352353 A	28/01/2009
		CN 101352353 B	19/10/2011
		CN 1479935 A	03/03/2004
		CN 1643641 C	20/07/2005
		CN 1809909 A	26/07/2006
		CN 1809909 B	16/11/2011
		CN 1833299 A	13/09/2006
		CN 1833299 B	16/06/2010
		CN 1992141 A	04/07/2007
		CN 1992141 B	16/10/2013
		EP 1328960 A1	23/07/2003
		EP 1476889 A1	17/11/2004
		EP 1627410 A2	22/02/2006
		EP 1636817 A2	22/03/2006
		JP 2004-511884 A	15/04/2004
		JP 2005-516343 A	02/06/2005
		JP 2006-524548 A	02/11/2006
		JP 2007-504636 A	01/03/2007
		KR 10-2003-0074605 A	19/09/2003
		KR 10-2004-0085163 A	07/10/2004
		TW 200421399 A	16/10/2004
		TW 200538721 A	01/12/2005
		TW I307110 A	01/03/2009
		US 2002-0094064 A1	18/07/2002
		US 2003-0142790 A1	31/07/2003
		US 2004-0028183 A1	12/02/2004
		US 2004-0114721 A1	17/06/2004
		US 2004-0213378 A1	28/10/2004
		US 2004-0240616 A1	02/12/2004
		US 2005-0281379 A1	22/12/2005
		US 2006-0008047 A1	12/01/2006
		US 2006-0018432 A1	26/01/2006
		US 2006-0274889 A1	07/12/2006
		US 2007-0009081 A1	11/01/2007
		US 2008-0043920 A1	21/02/2008
		US 6553096 B1	22/04/2003
		US 6850595 B2	01/02/2005
		US 6876724 B2	05/04/2005
		US 6980627 B2	27/12/2005
		US 7082182 B2	25/07/2006
		US 7085351 B2	01/08/2006
		US 7227924 B2	05/06/2007
		US 7359484 B2	15/04/2008
		US 7826595 B2	02/11/2010

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
		WO 02-031857 A1	18/04/2002
		WO 03-063195 A1	31/07/2003
		WO 2004-086439 A2	07/10/2004
		WO 2004-086439 A3	10/03/2005
		WO 2004-110111 A2	16/12/2004
		WO 2004-110111 A3	09/06/2005
		WO 2005-016113 A2	24/02/2005
		WO 2005-016113 A3	16/06/2005
		WO 2005-079246 A2	01/09/2005
		WO 2005-079246 A3	10/08/2006
WO 2014-028930 A1	20/02/2014	CN 104768467 A	08/07/2015
		US 2015-282774 A1	08/10/2015
US 2010-0239064 A1	23/09/2010	CN 101296658 A	29/10/2008
		CN 101296658 B	12/01/2011
		CN 101313214 A	26/11/2008
		CN 101313214 B	06/03/2013
		EP 1941264 A2	09/07/2008
		JP 2009-509580 A	12/03/2009
		US 2007-0053489 A1	08/03/2007
		US 7245692 B2	17/07/2007
		US 8155262 B2	10/04/2012
		WO 2006-116365 A2	02/11/2006
		WO 2006-116365 A3	23/08/2007
		WO 2007-038306 A2	05/04/2007
		WO 2007-038306 A3	25/10/2007
US 2012-0205549 A1	16/08/2012	CN 102597806 A	18/07/2012
		EP 2496964 A2	12/09/2012
		JP 2013-510423 A	21/03/2013
		WO 2011-055277 A2	12/05/2011
		WO 2011-055277 A3	29/12/2011
US 2009-0321643 A1	31/12/2009	CN 101490580 A	22/07/2009
		CN 101490580 B	04/07/2012
		EP 2047294 A2	15/04/2009
		EP 2047294 B1	20/11/2013
		JP 05313136 B2	09/10/2013
		JP 2009-545132 A	17/12/2009
		RU 2009105896 A	27/08/2010
		RU 2437119 C2	20/12/2011
		US 8193501 B2	05/06/2012
		WO 2008-012723 A2	31/01/2008
		WO 2008-012723 A3	27/03/2008