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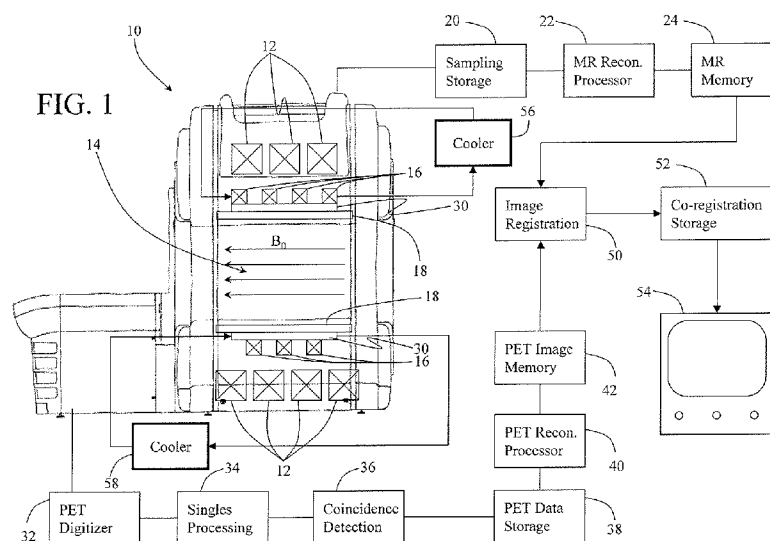
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(54) Title: THERMALLY STABILIZED PET DETECTOR FOR HYBRID PET-MR SYSTEM



(57) Abstract: In a hybrid PET-MR system, PET detector elements (30) are added in the bore (14), in close proximity to the gradient coils (16). Fluid coolant is supplied to transfer heat from the PET detector elements (30). Thermal insulation (80) insulates the fluid coolant and the PET detector elements (30) from the gradient coils (16). In some embodiments, a first coolant path (90) is in thermal communication with the electronics, a second coolant path (92) is in thermal communication with the light detectors, and a thermal barrier (94, 96) is arranged between the first and second coolant paths such that the first and second coolant paths can be at different temperatures (T_e , T_d). In some embodiments a sealed heat pipe (110) is in thermal communication with a heat sink such that working fluid in the heat pipe undergoes vaporization/condensation cycling to transfer heat from the detector elements to the heat sink.



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THERMALLY STABILIZED PET DETECTOR FOR HYBRID PET-MR SYSTEM DESCRIPTION

The present application relates to the diagnostic imaging arts. It finds particular application with hybrid positron emission tomography-magnetic resonance (PET-MR) systems where the PET detector elements are in close proximity to heat-generating elements of the MR system, and will be described with particular reference thereto. As the PET detector elements themselves also produce heat, the present application is not limited to hybrid systems, but can be used in standalone PET systems, PET/CT (computed tomography) systems, or nuclear medicine systems other than PET, such as single photon emission computed tomography (SPECT) systems.

In a bore-type MRI system, a large solenoidal main magnet produces a magnetic field along the axis of the bore. Generally, the closer that the main magnet is to the imaging region, the more cost effective is the design. Travelling inward from the main magnet are typically gradient coils, shim coils (sometimes built into the gradient coil assembly), a radio frequency (RF) shield, and RF coils. These coils, or MR bore components, take up more space. So even before a PET detector system is introduced into the bore of the MR system, space is at a premium. The PET detector is necessarily in close proximity to the MR bore components. PET-MR systems can also be built with a vertical field open MR system. The same space limitations apply to this type of system.

The gradient coils especially produce large amounts of heat. During MR scanning, the gradient coils dissipate on average about 15 kW with higher peak levels. This can cause the gradient coils to increase in temperature and water cooling may be used to limit their temperature to about 70° C. The RF coils also generate heat. RF coils and gradient coils also generate stray fields when pulsed and this time-varying stray field may interact with metal components of the PET detector, such as purpose-built RF shielding, and produce additional heat.

The PET detector elements, which generate heat on their own, are typically air cooled. As the light detectors are sensitive to heat (dark noise doubles about every 10° C increase in temperature), cooling to a temperature below the operating temperature of the gradient coils, to which they are in such close proximity, limits dark noise. But cooling in a PET-MR hybrid system is a problem because space is at such a premium.

The density of heat generating elements inside the main magnet is very high, while space available for cooling systems is minimal.

The present application provides a new and improved cooling system for PET elements within the bore of an MR apparatus that overcomes the above-referenced
5 problems and others.

In accordance with one aspect, a diagnostic imaging apparatus is provided. A magnetic resonance portion includes a main magnetic coil, a gradient coil assembly, and an RF coil assembly. The main magnetic coil generates a main magnetic field in an
10 imaging region of the apparatus. The gradient coil assembly overlays gradient magnetic fields on the main magnetic field. Shim coils may be located within and/or near the gradient coil assembly. Shim coils overlay correction magnetic fields to improve the homogeneity of the main magnetic field, inclusive of subject, *in vivo*, perturbations of the main magnetic field. The RF coil assembly at least transmits RF signals into the imaging
15 region. A positron emission tomography portion includes a fluid cooled detector array for detecting radiation indicative of nuclear decay events. The detector array is thermally proximate to the gradient coil assembly and is maintained at a lower temperature than the thermally proximate gradient coil assembly by fluid cooling of the fluid cooled detector array. At least one reconstruction processor produces an image representation from
20 detected signals originating from the imaging region. A display displays the reconstructed image representation.

In accordance with another aspect, a method of diagnostic imaging is provided. A main magnetic field is produced in an imaging region. Gradient fields are overlaid on the main magnetic field. Additional shim fields may also be overlaid.
25 Magnetic resonance is induced in a subject in the imaging region. The magnetic resonance is detected. A radiopharmaceutical is introduced into the subject, generally, sometime prior to the diagnostic imaging scan. Radiation indicative of radioactive decay events are detected with a detector array. The detector array is cooled with a fluid coolant. At least one of the detected magnetic resonance and the radiation is reconstructed into an image
30 representation of the subject. The image representation is displayed. It is to be appreciated that other orders for these steps are equally viable and have been contemplated.

In accordance with another aspect, a radiation detector is disclosed, comprising: scintillators; light detectors arranged to detect scintillations generated in the scintillators by interaction with radiation of interest; electronics configured to receive and process electrical signals indicative of radiation of interest received from the light detectors; a first coolant path in thermal communication with the electronics and configured to transmit flowing coolant fluid; a second coolant path in thermal communication with the light detectors and configured to transmit flowing coolant fluid; and a thermal barrier arranged between the first and second coolant paths such that the first and second coolant paths can be at different temperatures.

In accordance with another aspect, an imaging system is disclosed, comprising: a first modality imaging system defining a bore and configured to acquire first modality imaging data from inside the bore; and a positron emission tomography (PET) imaging system including a generally annular array of radiation detectors as set forth in the immediately preceding paragraph arranged coaxially with the bore of the first modality imaging system.

In accordance with another aspect, a radiation detector is disclosed, comprising: scintillators; light detectors arranged to detect scintillations generated in the scintillators by interaction with radiation of interest; electronics configured to receive and process electrical signals indicative of radiation of interest received from the light detectors; a heat sink; and a sealed heat pipe in thermal communication with the heat sink and the light detectors. The sealed heat pipe contains a working fluid and a wicking material or structure configured such that the working fluid undergoes vaporization/condensation cycling within the sealed heat pipe to transfer heat from the light detectors to the heat sink.

In accordance with another aspect, an imaging system is disclosed, comprising: a magnetic resonance scanner defining a bore and configured to acquire magnetic resonance data from inside the bore; and a positron emission tomography (PET) imaging system including a generally annular array of radiation detectors as set forth in the immediately preceding paragraph arranged coaxially with the bore to acquire PET data from inside the bore.

One advantage lies in improved cooling for PET detector elements.

Another advantage lies in compact cooling of the detector elements.

Another advantage lies in independent cooling systems for the gradient coils and the PET detector elements.

Still further advantages of the present invention will be appreciated to those of ordinary skill in the art upon reading and understand the following detailed description.

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

10 FIGURE 1 is a diagrammatic illustration of a combined PET/MR scanner, in accordance with the present application;

FIGURE 2 is a cross sectional view of a bore of the scanner of FIGURE 1, specifically showing a detector array between gradient and RF elements;

15 FIGURE 3 is a cross sectional view of a bore of the scanner of FIGURE 1, specifically showing a detector array between sections of a split gradient element;

FIGURE 4 is a diagrammatic illustration showing coolant flow directions through parallel flow pathways.

FIGURE 5 is a diagrammatic illustration showing a detector cooling configuration providing differential cooling for the electronics and the light detectors.

20 FIGURE 6 is a diagrammatic illustration showing a detector cooling configuration employing heat pipes.

FIGURE 7 is a diagrammatic illustration showing a planar view of the array of heat pipes of the detector cooling configuration of FIGURE 6, along with the connection with fluid-cooled heat sinks.

25 FIGURE 8 is a diagrammatic illustration showing a detector cooling configuration similar to that of FIGURE 6, but including thermal coupling to heat sinks at both ends of the heat pipes.

30 With reference to FIGURE 1, an embodiment of a combined or hybrid magnetic resonance (MR) and PET data acquisition system **10** is depicted. The combined or hybrid system **10** is capable of acquiring both MR data and PET data, for example from

MR and PET spatial regions that are at least partially overlapping (as illustrated) or spatially adjacent. It is to be understood that a multimodality system is ideal for temporally and spatially aligning or registering images; images taken from different scanners that are remotely located or spatially displaced are also contemplated, and similarly viable. The illustrated scanner **10** is an integrated system, but other scanner orientations are equally plausible, such as side-by-side systems, insert systems, and the like. The magnetic resonance scanner shown includes an annular magnet **12** including a plurality of conductor coil windings (diagrammatically depicted in FIGURE 1 by boxes with crossing lines) that generate a static magnetic field B_0 within an imaging region **14**. The magnet **12** may be superconducting or resistive in nature; in the former case the magnet **12** is typically disposed in a cryogenic dewar or other cooling system (not shown). The illustrated magnet **12** is a solenoidal magnet having a relatively horizontal B_0 field through the imaging region **14**. Though the polarity of the B_0 field is shown right to left, the opposite polarity is also suitable. In other embodiments, the magnet **12** may have a C or other configuration oriented to produce a vertical or otherwise oriented static magnetic field.

The magnetic resonance scanner also includes a magnetic field gradient assembly, which in the illustrative embodiment of FIGURE 1 includes gradient coil windings **16** that cooperatively superimpose magnetic field gradients on the static B_0 magnetic field responsive to selective energizing of selected gradient coil windings **16**. Optionally, the magnetic field gradient coil, magnet, or both may include other features not shown for forming, stabilizing, and dynamically adjusting the magnetic field, such as passive ferromagnetic shims, active shimming coils, and the like. The magnetic resonance scanner further includes a radio frequency excitation and reception system **18**. The radio frequency system includes at least one component, such as an illustrated whole body radio frequency coil **18**, that can be energized at a suitable radio frequency to excite magnetic resonance in a subject disposed in the imaging region **14**. The coil **18** can also operate as a radio frequency receiver to receive or detect magnetic resonance emanating from the imaging region **14** after RF excitation. In some embodiments, different coils are used for the excitation and reception operations. For example, the built-in coil **18** may be used to excite magnetic resonance, and a different, local coil or dedicated receive coil (not shown) may be positioned over, on, or close to the subject in the imaging region **14** to detect magnetic resonance. It is contemplated for the same magnetic resonance scanner to be

configurable in different ways using different combinations of built-in coils, local coils, or both.

Received magnetic resonance samples are stored in a magnetic resonance sampling storage **20**. A magnetic resonance reconstruction processor **22** applies a suitable reconstruction algorithm to reconstruct the magnetic resonance samples to form a reconstructed image that is stored in a magnetic resonance image memory **24**. The reconstruction processor **22** applies a reconstruction algorithm that comports with the selected spatial encoding used in generating the magnetic resonance data. For example, a Fourier transform reconstruction algorithm may be suitable for reconstructing Cartesian encoded magnetic resonance data. Optionally, part of the MR reconstruction may be done in-line, before data storage and completion of reconstruction.

With continuing reference to FIGURE 1, the illustrated combined or hybrid MR and PET data acquisition system **10** further includes radiation detectors for performing PET data acquisition. In the illustrative example of FIGURE 1, an annular array of radiation detectors **30** encircles the bore of the hybrid system. As will be described below, the illustrated array **30** includes a scintillator layer and a layer of electron multiplier based photon detectors, however, other detector configurations such as planar detector arrays disposed about the imaging region **14** are contemplated. Solid state radiation detectors and solid state optical detectors are also contemplated. The radiation detector array **30** is configured to detect 511 keV gamma rays that are emitted by positron – electron annihilation events. In PET data acquisition, two substantially simultaneous 511 keV gamma ray detection events are presumed to have originated from the same positron – electron annihilation event, which is located somewhere along a "line of response" (LOR) connecting the two substantially simultaneous 511 keV gamma ray detection events. This line of response is also sometimes called a projection or a ray, and the collected PET data is referred to as projection data.

In conventional PET, substantially simultaneous 511 keV gamma ray detection events are defined as two 511 keV gamma ray detection events occurring within a selected short time window, such as within four nanoseconds of each other. Any positron annihilation that does not occur at the center of the field of view (FOV) will have a small difference in time of arrival at the opposing detector elements, proportional to the difference in travel times for the gamma rays, about four nanoseconds at the edge of the

FOV. A related technique, called time-of-flight PET or TOF-PET, takes advantage of this small time difference to further localize the positron – electron annihilation event along the LOR with sub nanosecond precision.

The radiation detector array **30** of the hybrid system **10** is used to acquire
5 PET or TOF-PET data. The gamma ray detection events are processed by a PET digitization unit **32** that performs time-to-digital conversion (TDC) and analog-to-digital conversion (ADC) of detection events, and a singles processing unit **34** that performs clustering, energy estimation, time stamping, and positioning. The singles processing unit **34** optionally filters out detections that are outside of a selection energy window for the
10 expected 511 keV gamma ray energy. In some embodiments, the radiation detectors are pixelated. In other embodiments, clustering is applied by a block readout algorithm such as Anger logic or the like to provide further spatial localization refinement of the gamma ray detection events defining the projection. A coincidence detection processor **36** employs temporal windowing to identify gamma ray detection events that occurred
15 substantially simultaneously, and hence likely correspond to a common positron-electron annihilation event and hence define a projection or line of response.

For TOF processing, the time difference between the identified substantially simultaneous or coincident detection events is used to spatially estimate the positron-electron annihilation event along the line of response.

20 The resulting PET or TOF-PET data are stored in a PET data storage **38**. It is to be understood that the PET data could be stored before or after coincidence determination. A PET reconstruction processor **40** processes the projection or localized projection data using a suitable reconstruction algorithm to generate a reconstructed image that is stored in a PET images storage **42**. For example, maximum-likelihood expectation
25 maximization (ML-EM), a filtered backprojection algorithm or iterative reconstruction algorithm other than ML-EM can be employed. The system of FIGURE 1 includes a continuous radiation detector array **30** for PET that completely encircles the subject. Incomplete encirclement can lead to imaging artifacts due to "missing" projections or lines of response. For example, certain projections can be missed, and so information ordinarily
30 provided by such projections about related positions is unavailable. Advantageously, if time-of-flight PET data are acquired and reconstructed then the time-of-flight localization provides additional information that compensates for lost information even in the presence

of incomplete encirclement. There are, however, other ways of compensating for incomplete encirclement, and other detector configurations are certainly contemplated in the present application.

The MR and PET acquisitions are optionally performed concurrently.
5 Alternatively or additionally, MR and PET acquisition can be done sequentially (e.g., MR first followed by PET, or vice versa) or can be interleaved. An image registration processor **50** spatially registers and optionally temporally registers the reconstructed MR and PET images. If created, the co-registered images are stored in a co-registration image memory **52**. The images so registered are suitably displayed on a display device **54**,
10 rendered using suitable two- or three-dimensional rendering software, or otherwise processed.

A first cooling system **56** is provided to cool the gradient coils **16** to about 70° C or less when the gradient coils are operating at maximum average power level. The detector array **30** is thermally proximate to the gradient coils **16** in that the gradient coils
15 are close enough to the detector array **30** for heat from the gradient coils to transfer to the detector array **30**. A second cooling system **58** is provided to cool the detector array **30** to about 30° C or less, more desirably to about 20° C, in spite of the thermal proximity of the gradient coils **16**.

With reference now to FIGURE 2, an enlarged view of the imaging bore is provided. As mentioned previously, in one embodiment the detector array **30** extends 360°
20 about the longitudinal axis of the imaging bore. The same can be said for the RF coil **18**, the gradient coil **16**, and the main magnet **12**. In the embodiment of FIGURE 2, the detector array **30** is located between the gradient coils **16** and an RF shield **60** for the RF coils **18**. In a split gradient coil embodiment, as shown in FIGURE 3, the detector array **30**
25 can be located between portions of the gradient coil **16**. Other locations for the detector array **30** have also been contemplated, such as a removable system that is inserted and removed from the bore as needed.

In either of the embodiments of FIGURES 2 and 3, the detector array **30** is in close proximity to the gradient coils **16**. Active cooling keeps the detector array **30** from
30 reaching thermal equilibrium with the gradient coil **16**. In order to keep the detector array **30** at an acceptable operating temperature, the detector array **30** is provided with the closed circuit fluid cooling system **58**, as shown in FIGURE 4. FIGURE 4 is a cross-sectional,

close up view of the detector array **30**. The detector array **30** includes, among other things, a scintillation layer **62**, light detector elements **64**, and associated electronics **66**. The detector array **30** is all packaged within an RF shielded enclosure **68**. The enclosure **68** prevents excitation pulses from the RF coils **18** and magnetic resonance signals from the imaging region **14** from creating false signals in the detector array **30**. Also, the RF shielding prevents extraneous signals generated by the PET detector from being sensed by the MR RF coils. As mentioned before, all bore components, including the detector array **30** extend 360° around the bore.

The gradient coil temperatures range from room temperature, or slightly cooler, when not in operation, up to 70° C during operation. The surrounding bore space likewise rises in temperature. In addition, the electronics **66** of the PET detector generate heat. Stray field energy from pulsed MR fields deposited in surrounding RF shielding material also generates additional heat. Thus, to prevent all these heat sources from adversely affecting the operation of the detector array, the fluid cooling system **58** is applied to the PET detector array **30**. The cooling is applied to at least one surface of the detector. In one embodiment, the surface that is closest to the electronics and the gradient coils is fluid cooled and ends of the detector are insulated. In another embodiment, the circumferential surface and the end surfaces are all fluid cooled. In yet another embodiment, all accessible sides of the detector are fluid cooled, as shown in FIGURE 4. The cooling fluid is directed through conductive channels **70**. The channels **70** are made of a highly thermally conductive non-ferrous material, such as copper, aluminum, or stainless steel. As the detector array **30** itself is encased, the channels **70** that conduct heat away from the detectors are also enclosed by the RF shielding material **68**. Different, magnetically inert material may be used outside of the RF shield **68**. A feed channel **72** in particular may be made of or encased in a thermally insulating material, so that the coolant fluid does not draw heat away from components upstream of the detector array **30**, thereby becoming less effective in removing heat from the detector array **30** itself. In one embodiment, the feed channel runs between the RF screen **60** and the gradient coils **16**.

The cooling system is configured to provide efficient cooling of the PET detector **30** while conserving space. To this end, the cooling channels **70** are preferably thin, flat channels that are soldered or otherwise thermally connected directly to the circuitry **66**. It is also preferable that the channels **70** be added so as to not increase the

volume of the overall assembly, such as the channels **70** being arranged in complimentary fashions, such that facing circuit boards may fit together with little space wasted between them.

The cooling channels **70** about the detector array **30** are configured in such a manner to avoid creating closed conductive paths that would couple to the gradient fields. In one embodiment, this is accomplished by running the channels in zig-zag patterns primarily in an axial direction across the electronics carrying circuitry **66**. When the sides are cooled, the cooling channels can extend radially. Avoiding annular channels reduces eddy current generation. Apertures in the RF enclosure **68** are made for the inlet path **72** and an outlet path **74**. These apertures may be accomplished by using a waveguide. Dielectric breaks **76** in the cooling lines can be provided to break the channel conductive paths. A heat exchanger unit **78** provides flow and temperature control to the cooling fluid. The heat exchanger **78** can also chill the cooling fluid to below room temperature for more effective heat removal from the detector array **30**. The heat exchanger **78** itself may be air cooled within the imaging room or an adjacent room. Another option would be to interface the heat exchanger **78** with a larger cooling system, such as a general cooling system of the building. Heat from the coolant could be transferred to another fluid pathway and out to a rooftop air conditioning compressor, or the like.

In one embodiment, pure or deionized water with minimal conductive properties which has desirable thermal transport characteristics, is the coolant fluid. The rate of flow of the coolant is proportional to the total electronics power dissipation and inversely proportional to the outlet – inlet temperature difference. The flow path may include parallel paths to minimize the accumulated temperature rise per circulation pass and effect an overall pressure drop. A single path might accommodate lower volume and flow capabilities with increased pressure drop, but portions of the detector array **30** at the end of the flow path may not be cooled as well as portions at the beginning of the flow path, meaning that the detector components further downstream might not be cooled to the same temperature as the upstream components. This may lead to, among other things, non-uniform image artifacts. Local provisions, such as heat sinks, can provide heat conduction paths away from the electronics **66** to the fluid channels **70**. Alternatively, other coolant fluids, such as refrigerants, liquid nitrogen, forced air, or so forth could be used as the fluid coolant in the channels **70**, and it is to be understood that “fluid” as used in the present

application includes all of these. When using a colder than room temperature coolant, air flow can be provided through the bore to help prevent condensation.

In an alternate embodiment, a thermally conductive dielectric oil could be used as the coolant. In this embodiment, the dielectric coolant could come into direct
5 contact with the electronics **66** and the detector elements instead of indirectly via the channel walls. In this immersion-cooling embodiment, the entire detector array **30** is encased in a fluid tight containment chamber, such as a cylindrical or elliptically profiled envelope. This method would obviate many fluid channels and heat sinks. Baffles in the containment chamber can be included to reduce the potential for non-uniform flow rates
10 and hot spots.

In another alternate embodiment, Peltier cooling is used to cool the detector array **30**. This could be used as a stand-alone method, or selective Peltier cooling could be used to cool trouble spots as a supplement to one or more of other methods described above. In the split gradient coil configuration of FIGURE 3, it is advantageous for the
15 cooler **56** and associated gradient cooling lines to be arranged to cool the split gradient coil from the inside to the outside, that is, from the region proximate to the gap of the split gradient coil to the outer extremities of the split gradient coil. This is advantageous because the gradient coil will then be coolest and have most stable temperature near the gap proximate to the detector array **30**. Optionally, the cooling lines are also concentrated near
20 the gap to further enhance gradient coil cooling efficiency nearest the detector array **30**.

In another alternate embodiment, a common fluid cooling system is used to cool both the gradient coils **16** and the detector array **30**. One advantage of this embodiment would be savings in both cost and space. A disadvantage, however, would lie in the fact that the temperature of the detector array **30** might be influenced by the gradient
25 coil **16** temperature. The cooling fluid could cool the detector array to the lower temperature with the cooling fluid leaving the detector array going to cool or help cool the gradient coils **16**. Also, separate circuits could be utilized, so that only a portion of the heat exchanger **78** is shared. In a dual cooling system environment, insulation **80** can be used, at least in part, to thermally isolate the gradient coils **16** from the detector array **30**. This
30 way, independent benchmarks can be set for each cooling system. With a combined cooling system, the gradient coils **16** could be cooled to the desired operating temperature of the detector array **30**, which would take significantly more rigorous cooling, due to the

greater thermal mass of the gradient coil **16**. Another alternative is to have a combined cooling system with separate heat exchangers for the PET detector array and other elements that require cooling.

With reference to FIGURE 5, a detector cooling configuration is shown which provides differential cooling for the electronics **66** and the light detectors **64**. The electronics may include, for example, time-to-digital converter (TDC) elements, analog-to-digital converter (ADC) elements, field-programmable gate array (FPGA) or other logic elements, low drop-out (LDO) regulators, and so forth. These components generate a substantial amount of heat that is to be removed, but performance of the electronics **66** is generally not strongly correlated with temperature so long as the electronics **66** are maintained at a suitably low operating temperature, denoted T_e in FIGURE 5. Thus, a first cooling path **90** proximate to and cooling the electronics **66** is configured to provide a relatively high coolant fluid flow rate, denoted F_e in FIGURE 5, to provide high heat transfer.

The light detectors **64**, such as silicon photomultiplier (SiPM) elements or photomultiplier tubes, have operating characteristics which are strongly temperature dependent. For example, for some SiPM detectors the gain increases by about 5-10% for each temperature increase of 1 °C. However, the light detectors **64** generally output substantially less heat than the electronics **66**. Thus, a second cooling path **92** proximate to and cooling the light detectors **64** is configured to provide a relatively lower flow rate, denoted F_d in FIGURE 5, which maintains the light detectors **66** at a precise temperature T_d that does not vary substantially with time.

The electronics **66** and associated cooling path **90** on the one hand, and the light detectors **64** and associated cooling path **92**, on the other hand, are arranged to be substantially thermally isolated from each other. This can be accomplished by passive thermal insulation material **94**, such as an insulating dielectric material or an air gap, disposed between the electronics **66** and associated cooling path **90** on the one hand, and the light detectors **64** and associated cooling path **92**, on the other hand. Additionally or alternatively, this can be accomplished by active thermal isolation provided by an isolation cooling path **96** flowing coolant fluid at a flow rate F_i .

The scintillators **62** generally do not require cooling. However, the scintillators **62** are arranged in close proximity to the light detectors **64**, which must be kept

at a stable temperature. Accordingly, in the embodiment of FIGURE 5 additional passive thermal insulation material **98** is disposed around the scintillators **62** so that the passive insulation **94**, **98** substantially contains a stable-temperature region including the light detectors **64**. The detector temperature management system as a whole is suitably contained in an outer container **100** surrounding the detector module, with a coolant fluid inlet **102** at one side distributing coolant fluid flows F_e , F_d , F_i into the cooling paths **90**, **92**, **96**, and a coolant fluid outlet **104** at the opposite side draining the coolant fluid flows F_e , F_d , F_i from the detector module. The differing relative flow rates for the coolant fluid flows F_e , F_d , F_i are suitably defined by fluid flow resistances (or, equivalently, fluid flow conductances) of the corresponding cooling paths **90**, **92**, **96**. Optionally, one or more temperature sensors **106** are disposed to measure the temperature of the light detectors **64**, and the flow into the inlet **102** (or, alternatively, out of the outlet **104**) is controlled to maintain the temperature sensor **106** at a desired temperature reading. Alternatively, the temperature sensor can be arranged to measure the temperature T_d of the coolant fluid path **92** that cools the light detectors **64**, as this temperature T_d should be close to the temperature of the light detectors **64**. Controlling coolant fluid flow based on the detector temperature (or, approximately, based on the temperature T_d) is advantageous because this is the critical temperature that should be maintained stable.

Although not illustrated, it is also contemplated to provide active, e.g. externally controllable or temperature feedback-controlled, valving to adjust the relative flow rates F_e , F_d , F_i into the cooling paths **90**, **92**, **96**. Additionally, although not shown it is contemplated to provide passive thermal insulation material surrounding the electronics **66** and associated flow path **90**.

In another contemplated variation, The cooling of the module as shown in FIGURE 5 is modified by using a counter-current setup for each individual plane to reduce temperature gradients within the module. This can be accomplished by placing the inlet **102** and outlet **104** on the same side of the detector module. It is expected that use of a counter-current setup will reduce temperature gradients by 1-2 orders of magnitude which can be expected to improve temperature stability. Alternatively, the cooling can be cascaded as the actual heat generated by the light detectors **64** is estimated to be 100 times lower than the heat generated by the electronics **66**. In either a counter-current or cascaded design, it is advantageous to first cool the light detectors **64** and then the electronics **66**, for

example by flowing the coolant first through the cooling path **92** proximate to the light detectors **64** and then through the cooling path **90** proximate to the electronics **66**. This arrangement of coolant fluid flow promotes a precise and lower temperature T_d proximate to the light detectors **64** as compared with the temperature T_e proximate to the electronics **66**. The temperature T_e of the electronics **66** is where most of the heat is dissipated, but the electronics temperature does not have to be as accurate and it can be cooled by the coolant fluid after passing proximate to the light detectors **64**.

As yet another variation on the design of FIGURE 5, with reference to inset **II** shown in FIGURE 5 micro-Peltier elements **108** can provide an active, controllable thermal resistance between the cooling path **92** and the light detectors **64**. As an example, standard Peltier elements are suitably based on Bismuth Telluride (Bi_2Te_3) and Bismuth Selenide (Bi_2Se_3). An arrangement similar to that used to cool semiconductors using silicon-based Peltier elements (described, for example, in Mathews et al., U.S. Pat. No. 6,800,933 which is incorporated herein by reference) can be used, in which micro-Peltier devices are disposed in non-sensitive areas on top of semiconductor-on-insulator (SOI) substrates used in the light detectors **64**. It is possible to implement these structures also in a separate device mounted on back of the light detectors **64**. In this approach, local temperature gradients can be tuned and controlled to within about 0.1°C or less, so as to ensure exact and constant temperature at each light detector element. In these embodiments, the temperature of the cooling medium does not have to be as low as the light detector temperature, eliminating condensation and expensive cooling of the medium below room temperature.

The radiation detector system of FIGURE 5 employs coolant fluid. The coolant fluid may be, for example, deionized water, forced air, a refrigerant, or so forth. In such embodiments, the inlet and outlet **102**, **104** are suitably connected with external supply and return lines (e.g., lines **72**, **74** of FIGURE 4), for example using a permanent connection such as a soldered connection, or a detachable connector such a threaded coupling or a quick-disconnect connection.

With reference to FIGURES 6 and 7, in another embodiment one or more heat pipes, such as an illustrated array of parallel linear heat pipes **110**, are used to provide heat transfer from the components of a radiation detector module to one or more heat sinks **112** made of a thermally conductive material such as a thermally conductive ceramic,

aluminum oxide, aluminum nitride, or so forth. A thermally conductive metal such as copper can also be used for the heat sinks **112**, but should be laminated or otherwise configured to suppress eddy current formation. Optionally, a cooling fluid pathway **114** passes through the heat sinks **112** to provide removal of heat transferred into the heat sinks **112** by the heat pipes **110**.

Each heat pipe **110** includes a sealed interior volume **120** containing at least one working fluid, such as water, a solvent such as alcohol, or so forth, or a mixture of working fluids. A wicking material or structure **122** consisting of a porous material, grooves formed on the surface, or another material or structure supportive of capillary or wicking action is disposed on interior surfaces of the interior volume **120**.

In operation, the working fluid vaporizes proximate to the end **124** of the heat pipe **110** distal from the heat sink **112**, or at other "hot spots" along the heat pipe **110**. A heat of vaporization is transferred from the hot spot to the vaporized working fluid, which is diagrammatically indicated by arrows **126**. The vaporized working fluid **126** moves toward the heat sink **112** due to vapor concentration gradients. The vaporized working fluid **126** condenses proximate to the end **128** of the heat pipe **110** proximate to the heat sink **112**. Condensation of the working fluid at the proximate end **128** transfers a heat of condensation from the condensing working fluid to the heat sink **112**. The condensed working fluid is drawn back toward the distal end of the heat pipe **110** by capillary or wicking action associated with the wicking material or structure **122**. When the working fluid vaporizes again at a hot spot, the heat transfer cycle is complete. Advantageously, the condensed fluid return by capillary forces in the wicking material or structure **122** is independent of spatial position of the heat pipe **110**.

For use in a hybrid system including an MR component, the heat pipes **110** are preferably made of non-magnetic and electrically non-conductive materials, such as high thermal conductivity ceramics such as aluminum nitride or alumina. If metallic or other electrically conductive materials are used, they should be laminated, arranged as a composite in an electrically non-conductive matrix material, or otherwise configured to suppress eddy current formation. As illustrated in FIGURES 6 and 7, the array of heat pipes **110** is generally planar and is disposed between the light detectors **64** and the associated electronics **66** to cool both. Since the vaporized working fluid **126** is transported

rapidly, there is an intrinsic temperature isolation between the relatively hotter electronics **66** and the relatively cooler light detectors **64**.

Advantageously, the connection of the heat pipes **110** and the heat sinks **112** can be by thermally conductive surface contacts **130**, optionally enhanced by a pressure fitting, threaded connection, or so forth, and does not entail flow of cooling fluid between the heat sinks **112** and the heat pipes **110** and hence does not include fluid connections therebetween. Advantageously, the heat pipes **110** are permanently sealed units that are not opened when the radiation detector module is mounted or dismounted.

Another advantage is that the temperature at the exterior of the heat pipes **110** is substantially controlled by the vaporization temperature of the working fluid and is substantially constant along the heat pipes **110**. The vaporization temperature is a function of the cooling medium or media.

With reference to FIGURE 8, to further reduce temperature gradients, it is contemplated to have each heat pipe **110** in thermal communication with more than one heat sink **112**, such as having heat sinks **112** at both ends of the heat pipe **110** as shown in FIGURE 8. It is also contemplated to arrange an array of crossing linear heat pipes, or to use a generally planar heat pipe, or to employ other heat pipe geometries to enhance heat removal and temperature stability.

The embodiment of FIGURE 8 also includes the light detectors **64** and the electronics **66** integrated as a single layer, for example monolithically integrated on or in a single common layer of silicon substrate material. See, for example, WO 2006/111883 A2 (published Oct. 26, 2006 and incorporated herein by reference in its entirety) which relates to a digital silicon photomultiplier (SiPM) for time-of-flight PET in which the digital SiPM includes a detector arrays layer and a buried CMOS digital processing circuitry layer monolithically formed in or on a common silicon substrate. In the embodiment of FIGURE 8, the heat pipe **110** has wicking material or structure **122** disposed on the interior surfaces of the interior volume **120** both distal from and proximate to the cooled elements **64**, **66**. However, since in the embodiment of FIGURE 8 the cooled elements **64**, **66** are disposed together on one side of the heat pipe **110**, it is contemplated to dispose the wicking material or structure only on the interior surface proximate to the cooled elements **64**, **66**.

As another variation, although in the embodiments illustrated in FIGURES 6-8 the heat pipes **110** are in direct contact or in close proximity to the cooled elements **64**,

66, in other embodiments the heat pipes may be disposed in a block of thermally conductive ceramic material or electrically non-conductive composite material that in turn is in intimate thermal contact with the cooled elements 64, 66.

Although described in the illustrative context of a hybrid imaging system incorporating magnetic resonance imaging, it is to be appreciated that the radiation detector module cooling configurations disclosed with reference to FIGURES 4-8 are also contemplated for use in other imaging systems employing radiation detectors, such as stand-alone PET imaging systems, PET/CT (computed tomography) imaging systems, gamma cameras, and so forth.

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

CLAIMS

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

1. A diagnostic imaging apparatus comprising:
 - a magnetic resonance portion that includes:
 - a main magnetic coil assembly (12) for generating a main magnetic field in an imaging region (14) of the apparatus;
 - a gradient coil assembly (16) for overlaying gradient magnetic fields on the main magnetic field;
 - an RF coil assembly (18) for at least transmitting RF signals into the imaging region;
 - a positron emission tomography portion including:
 - a fluid cooled detector array (30) for detecting radiation indicative of nuclear decay events, the detector array being thermally proximate to the gradient coil assembly and maintained at a lower temperature than the thermally proximate gradient coil assembly by fluid cooling of the fluid cooled detector array;
 - at least one reconstruction processor (22, 40) for producing an image representation from detected signals originating from the imaging region; and
 - a display (54) for displaying the reconstructed image representation.
2. The diagnostic imaging apparatus as set forth in claim 1, further including:
 - an RF shield enclosure (68) that encases the detector array and shields the array from RF signals.
3. The diagnostic imaging apparatus as set forth in claim 1, wherein the fluid coolant includes one of water and dielectric oil.
4. The diagnostic imaging apparatus as set forth in claim 1 wherein a coolant loop (58) includes a cylindrical envelope.

5. The diagnostic imaging apparatus as set forth in claim 1, wherein a coolant loop (58) includes a plurality of channels extending generally parallel to the main magnetic field.

6. The diagnostic imaging apparatus as set forth in claim 4, further including:
at least one dielectric break (76) in the coolant loop that disrupts eddy current paths.

7. The diagnostic imaging apparatus as set forth in claim 1, further including:
a closed circuit coolant loop (58) that encases the detector array, the detector array being immersed in thermally conductive dielectric oil.

8. The diagnostic imaging apparatus as set forth in claim 1, wherein the detector array (30) includes:

a generally annular array of radiation detectors (64) extending around the imaging region;

detector electronics (66) disposed between the generally annular array (64) and the gradient coil assembly (16);

cooling fluid paths defined at least between the detector electronics (66) and the gradient coil assembly (16).

9. The diagnostic imaging apparatus as set forth in claim 8, further including:
thermal insulation (80) disposed between the cooling fluid paths and the gradient coil array (16).

10. The diagnostic imaging apparatus as set forth in claim 9, wherein the cooling fluid paths (70) and the thermal insulation (80) are disposed in a generally annular region around the detector electronics (66).

11. The diagnostic imaging apparatus as set forth in claim 10, wherein the thermal insulation (80) extends radially along sides of the detector electronics (66) and the array of radiation detectors (64).

12. The diagnostic imaging apparatus as set forth in claim 10, wherein the detector array is located between the gradient coil assembly (16) and the RF coil assembly (18) and an RF screen (60) extends between the array of radiation detectors (64) and the RF coil (18).

13. The diagnostic imaging apparatus as set forth in claim 12, wherein the detector array (30) is located between gradient coils (16) in a split gradient coil system.

14. The diagnostic imaging apparatus as set forth in claim 10, wherein the cooling fluid paths (70) further extend radially along the sides of the detector electronics (66) and array of detector elements (64).

15. The diagnostic imaging apparatus as set forth in claim 8, wherein the cooling fluid paths (70) include a first cooling path (90) arranged to cool the detector electronics (66) and a second cooling path (92) arranged to cool the detector elements (64).

16. The diagnostic imaging apparatus as set forth in claim 15, wherein first and second coolant paths (90, 92) are driven by a common inlet (102) and a common outlet (104) in one of (i) a parallel arrangement, (ii) a counter-current arrangement, or (iii) a cascaded arrangement in which coolant fluid passes through the second cooling path (92) and then into the first cooling path (90).

17. The diagnostic imaging apparatus as set forth in claim 15, further comprising:
a temperature sensor (106) operatively coupled with at least one of the second coolant path (92) and the detector elements (64), a fluid flow (F_d) through the second coolant path being controlled to maintain a stable temperature as indicated by the temperature sensor.

18. The diagnostic imaging apparatus as set forth in claim 15, further comprising:
Peltier elements (108) disposed in a thermal pathway between the second cooling path (92) and the detector elements (64).

19. The diagnostic imaging apparatus as set forth in claim 8, further including:
one or more heat pipes (110) in thermal communication with the radiation detectors (64) to provide a stable temperature for the radiation detectors (64).
20. The diagnostic imaging apparatus as set forth in claim 8, wherein the array of radiation detectors (30) includes one of:
an array of solid state detector elements (64); or
an array of scintillators (62) optically coupled with an array of light detectors (64).
21. The diagnostic imaging apparatus as set forth in claim 1, further including:
a common coolant circuit that cools both the gradient coil (16) and the detector array (30).
22. A method of diagnostic imaging comprising:
detecting the magnetic resonance using a magnetic resonance scanner;
detecting radiation indicative of radioactive decay events with a detector array (30) in thermal proximity with the magnetic resonance scanner;
cooling the detector array (30) with a fluid coolant;
reconstructing at least one of the detected magnetic resonance and the radiation into an image representation of the subject; and
displaying the image representation.
23. The method as set forth in claim 22, further including:
shielding the detector array (30) with an RF shield (68) that shields the array (30) from RF signals.
24. The method as set forth in claim 22, wherein the step of cooling the detector array (30) with a fluid coolant includes:
pumping the coolant through multiple channels (70) adjacent to the detector array (30).

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

thermally isolating at least one portion of a coolant loop (58) from gradient coils (16) that overlay the gradient fields.

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

cooling detector electronics (66) of the detector array (30) with the fluid coolant;

thermally isolating the detector array (30) from a gradient coil assembly (16) that overlays the gradient fields on the main magnetic field.

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

cooling gradient coils (16) with the fluid coolant.

28. The method as set forth in claim 22, wherein the cooling comprises:

cooling electronics (66) of the detector array with a first cooling path (90) through which fluid coolant flows; and

cooling radiation detectors (64) of the detector array with a second cooling path (92) different from the first cooling path through which fluid coolant flows;

wherein the first and second cooling paths (90, 92) both receive fluid coolant directly or indirectly from a common fluid coolant inlet (104).

29. The method as set forth in claim 28, wherein the first and second cooling paths (90, 92) are cascaded with fluid coolant flowing through the second coolant path (92) to cool the radiation detectors (64) and then through the first coolant path (90) to cool the electronics (66).

30. The method as set forth in claim 22, wherein the cooling comprises:

disposing at least a portion of the fluid coolant in a sealed heat pipe (110) also containing a wicking material or structure (122), the sealed heat pipe (110) operatively coupled with the detector array (30) and with at least one heat sink (112) such that vaporization/condensation cycling of the fluid coolant in the sealed heat pipe transfers heat from the detector array to the heat sink.

31. The method as set forth in claim 22, further comprising:

controlling thermal resistance between the detector array **(30)** and the fluid coolant using micro-Peltier elements **(108)**.

32. An imaging system comprising:

a positron emission tomography portion;

a magnetic resonance portion;

a cooling system that shares at least one component for cooling the positron emission tomography portion and the magnetic resonance portion.

33. The imaging system as set forth in claim 32, wherein the cooling system includes a coolant that is one of a liquid and a refrigerant.

34. The imaging system as set forth in claim 33, wherein the at least one component includes shared coolant.

35. The imaging system as set forth in claim 32, wherein the at least one component includes a shared heat exchanger **(78)**.

36. The imaging system as set forth in claim 32, wherein the at least one component includes shared heat sinking material.

37. An imaging system comprising:

a magnetic resonance portion;

a positron emission tomography portion located within a bore of the magnetic resonance portion;

a magnetic resonance shield **(68)** that encases the positron emission tomography portion

a fluid cooling system that is introduced into the magnetic resonance shield **(68)** to cool at least a detector array **(30)** of the positron emission tomography portion.

38. The imaging system as set forth in claim 37, wherein the fluid cooling system cools the detector array (30) and gradient coils (16) of the magnetic resonance portion.

39. A radiation detector comprising:

scintillators (62);

light detectors (64) arranged to detect scintillations generated in the scintillators by interaction with radiation of interest;

electronics (66) configured to receive and process electrical signals indicative of radiation of interest received from the light detectors;

a first coolant path (90) in thermal communication with the electronics and configured to transmit flowing coolant fluid;

a second coolant path (92) in thermal communication with the light detectors and configured to transmit flowing coolant fluid; and

a thermal barrier (94, 96) arranged between the first and second coolant paths (90, 92) such that the first and second coolant paths can be at different temperatures (T_e , T_d).

40. The radiation detector as set forth in claim 39, wherein the thermal barrier includes at least one of (i) passive thermal insulation material (94) and (ii) an isolation cooling path (96) flowing coolant fluid.

41. The radiation detector as set forth in claim 39, wherein the second coolant path (92) is maintained at a relatively lower temperature (T_d) that is substantially stable, and the first coolant path (90) is maintained at a relatively higher temperature (T_e).

42. The radiation detector as set forth in claim 39, further comprising:

micro-Peltier elements (108) disposed between the second cooling path (92) and the light detectors (64).

43. An imaging system comprising:

a first modality imaging system (10) defining a bore (14) and configured to acquire first modality imaging data from inside the bore; and

a positron emission tomography (PET) imaging system including a generally annular array of radiation detectors as set forth in claim 39 arranged coaxially with the bore of the first modality imaging system.

44. A radiation detector comprising:

scintillators **(62)**;

light detectors **(64)** arranged to detect scintillations generated in the scintillators by interaction with radiation of interest;

electronics **(66)** configured to receive and process electrical signals indicative of radiation of interest received from the light detectors;

a heat sink **(112)**; and

a sealed heat pipe **(110)** in thermal communication with the heat sink and the light detectors, the sealed heat pipe containing a working fluid and a wicking material or structure **(122)** configured such that the working fluid undergoes vaporization/condensation cycling within the sealed heat pipe to transfer heat from the light detectors to the heat sink.

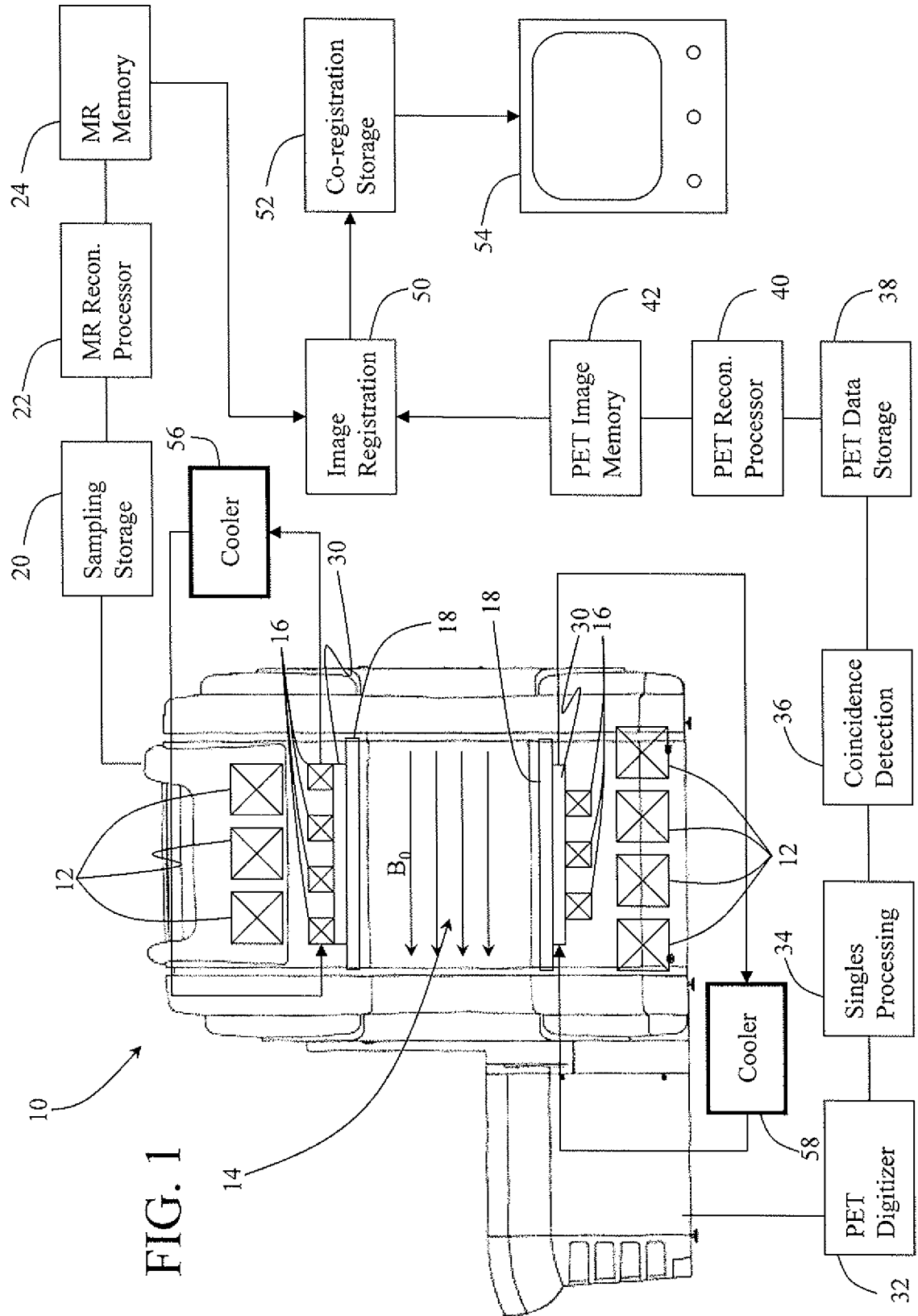
45. The radiation detector as set forth in claim 44, wherein the sealed heat pipe **(110)** is in thermal communication with the electronics **(66)** such that the working fluid undergoing vaporization/condensation cycling within the sealed heat pipe transfers heat from the electronics to the heat sink.

46. The radiation detector as set forth in claim 44, wherein the radiation detector is configured as a module including the scintillators **(62)**, the light detectors **(64)**, the electronics **(66)**, and the heat pipe **(110)**, and the module is operatively connectable with at least a portion of the heat sink **(112)** by a thermally conductive surface contact **(130)** that does not entail flow of cooling fluid between the heat sink and the heat pipe.

47. An imaging system comprising:

a magnetic resonance scanner **(10)** defining a bore **(14)** and configured to acquire magnetic resonance data from inside the bore; and

a positron emission tomography (PET) imaging system including a generally annular array of radiation detectors as set forth in claim 44 arranged coaxially with the bore to acquire PET data from inside the bore.



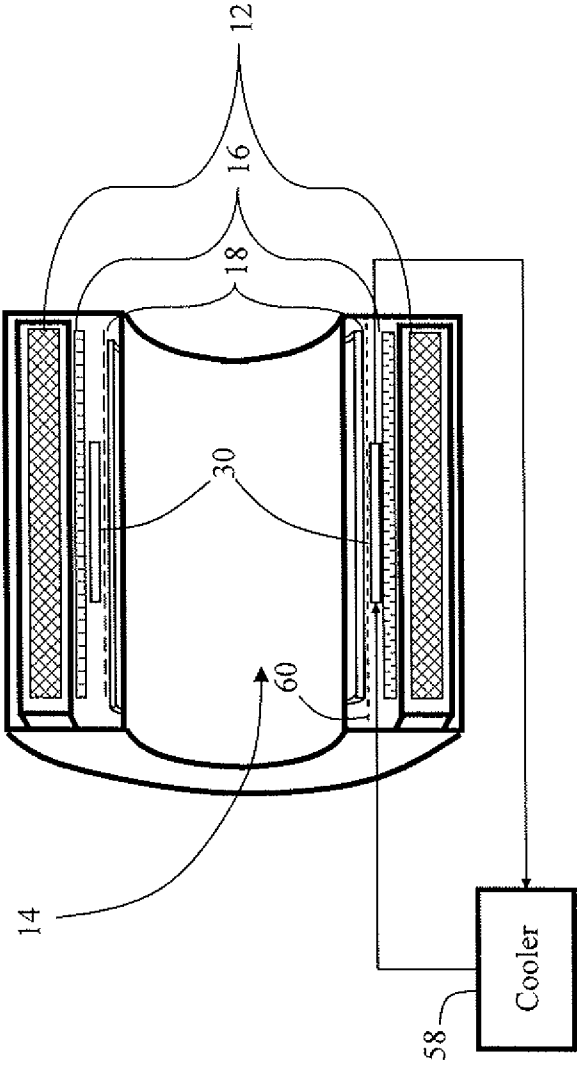


FIG. 2

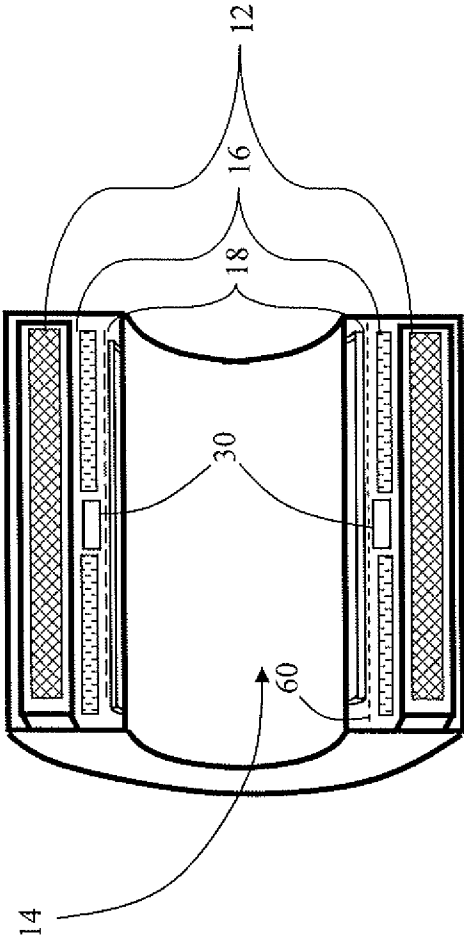


FIG. 3

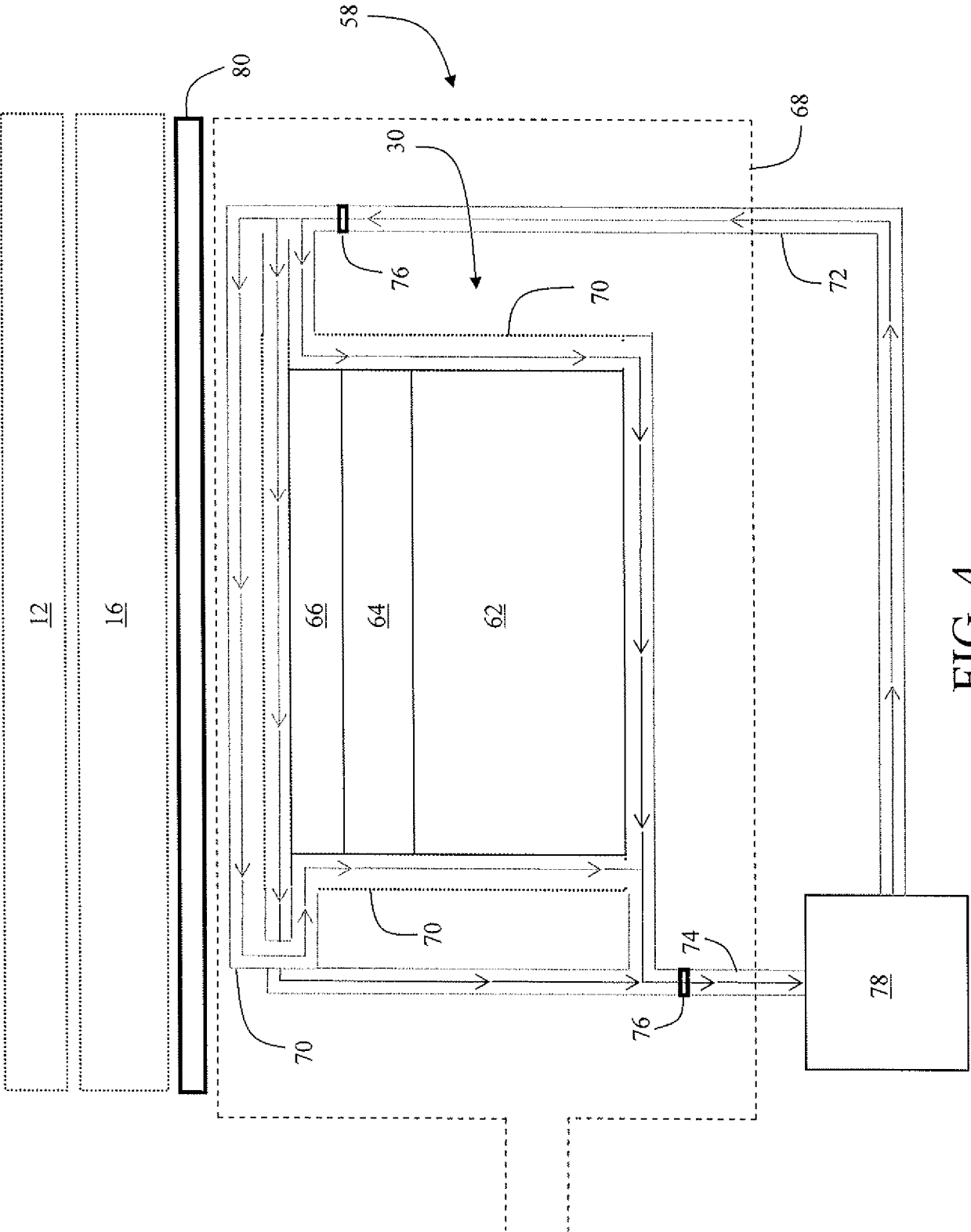


FIG. 4

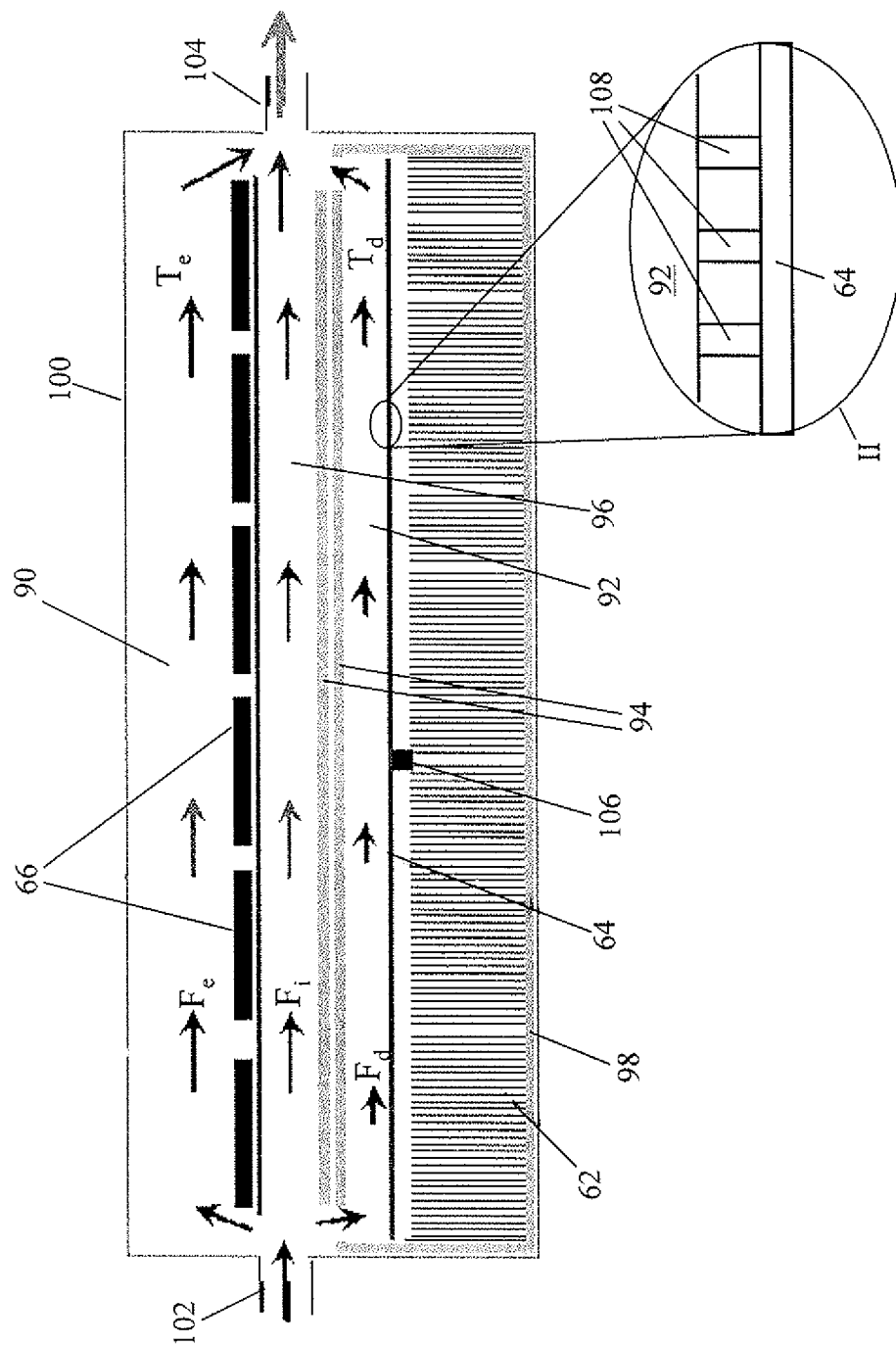


FIG. 5

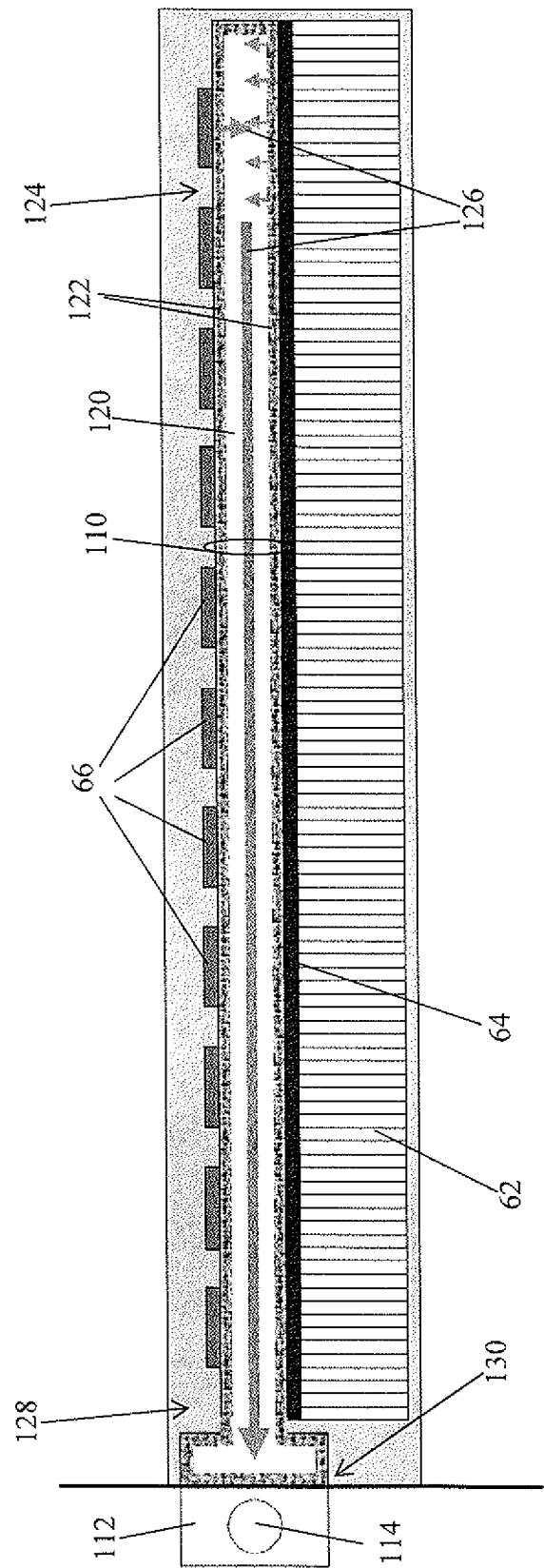


FIG. 6

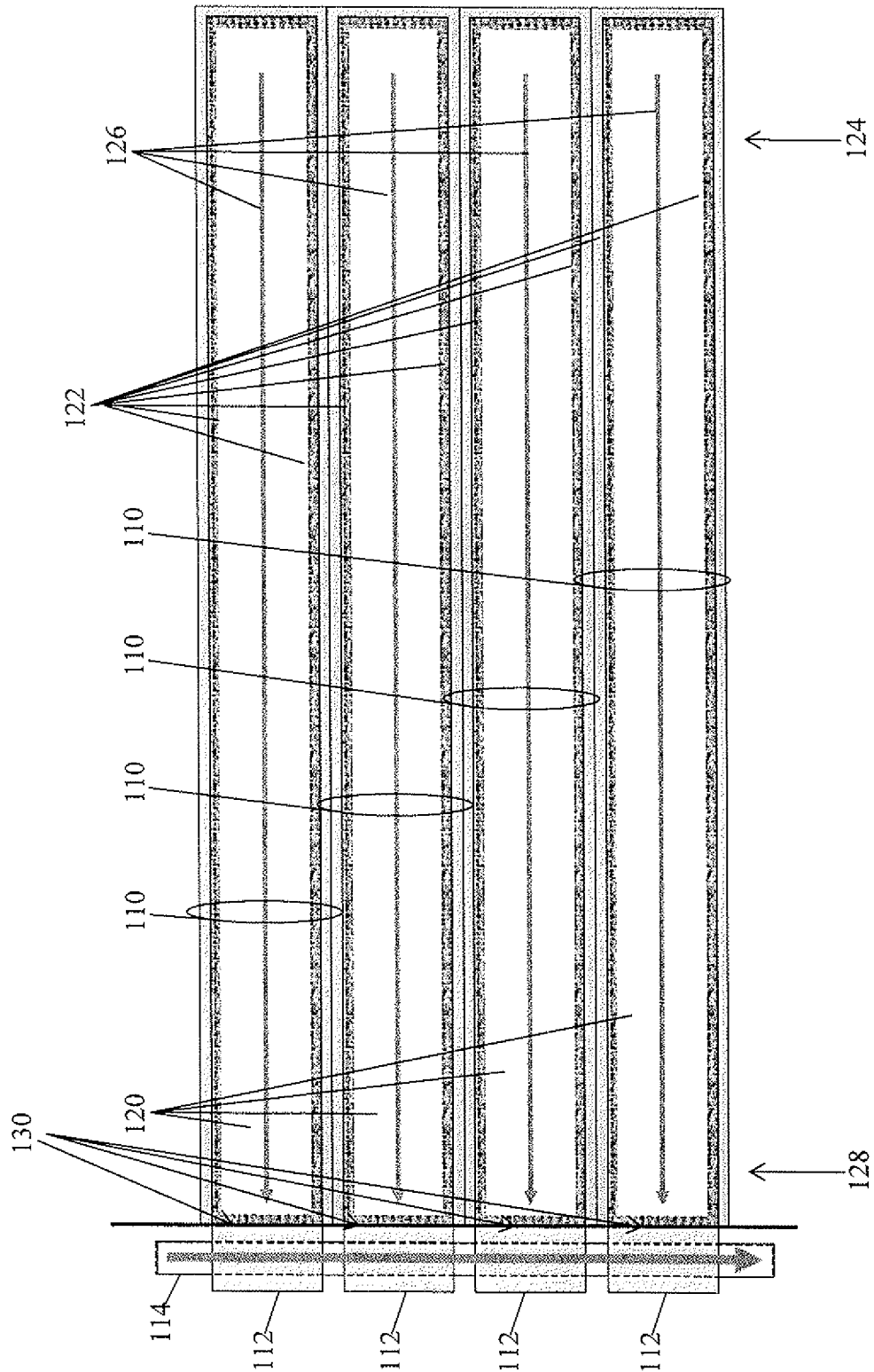


FIG. 7

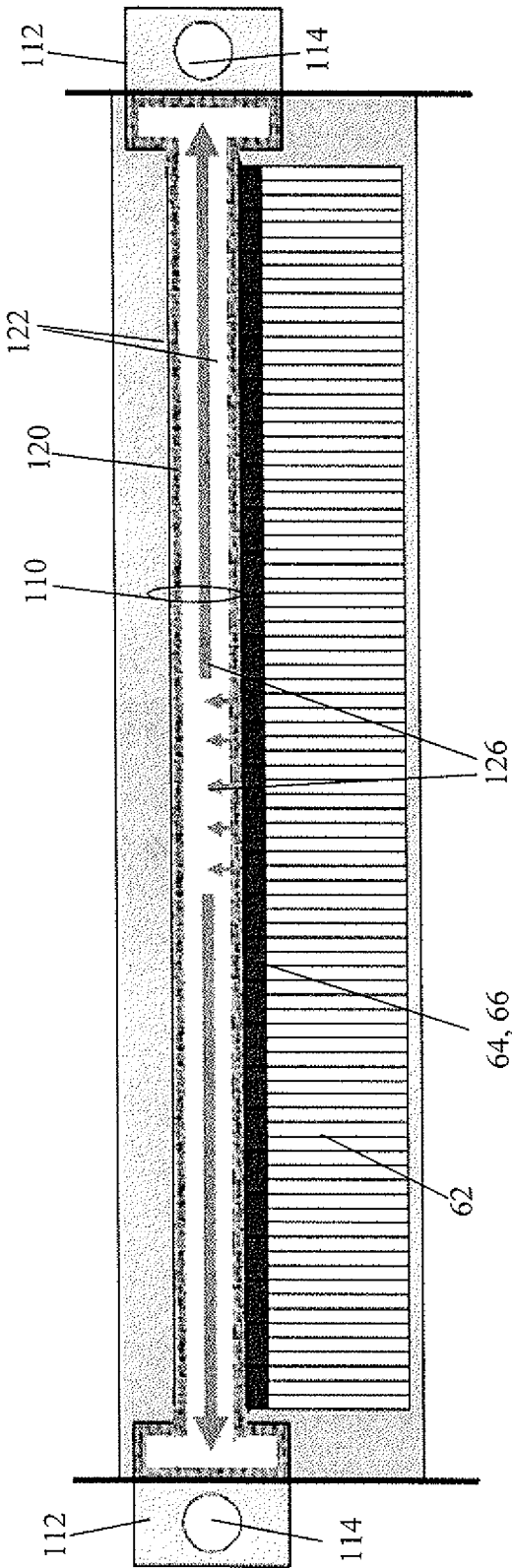


FIG. 8