

US 20020024024A1

(19) **United States**(12) **Patent Application Publication**
Mastrippolito et al.(10) **Pub. No.: US 2002/0024024 A1**(43) **Pub. Date: Feb. 28, 2002**(54) **DEVICE FOR NON-INVASIVE ANALYSIS BY
RADIO-IMAGING, IN PARTICULAR FOR
THE IN VIVO EXAMINATION OF SMALL
ANIMALS, AND METHOD OF USE****Related U.S. Application Data**(62) Division of application No. 08/981,387, filed on Apr.
13, 1998, now Pat. No. 6,225,631.(76) Inventors: **Roland Mastrippolito**, Montigny Le
Bretonneux (FR); **Lydie Ploux**, Le
Mans (FR); **Pierre Yves Charon**,
Gif-Sur-Yvette (FR); **Laurent Pinot**,
Lardy (FR); **Luc Valentin**,
Bures-Sur-Yvette (FR); **Alejandro**
Anibal Valda Ochoa, Villalba (ES);
Rainer Siebert, Port Marly (FR);
Philippe Laniece, Paris (FR); **Herve**
Tricoire, Palaiseau (FR)(30) **Foreign Application Priority Data**

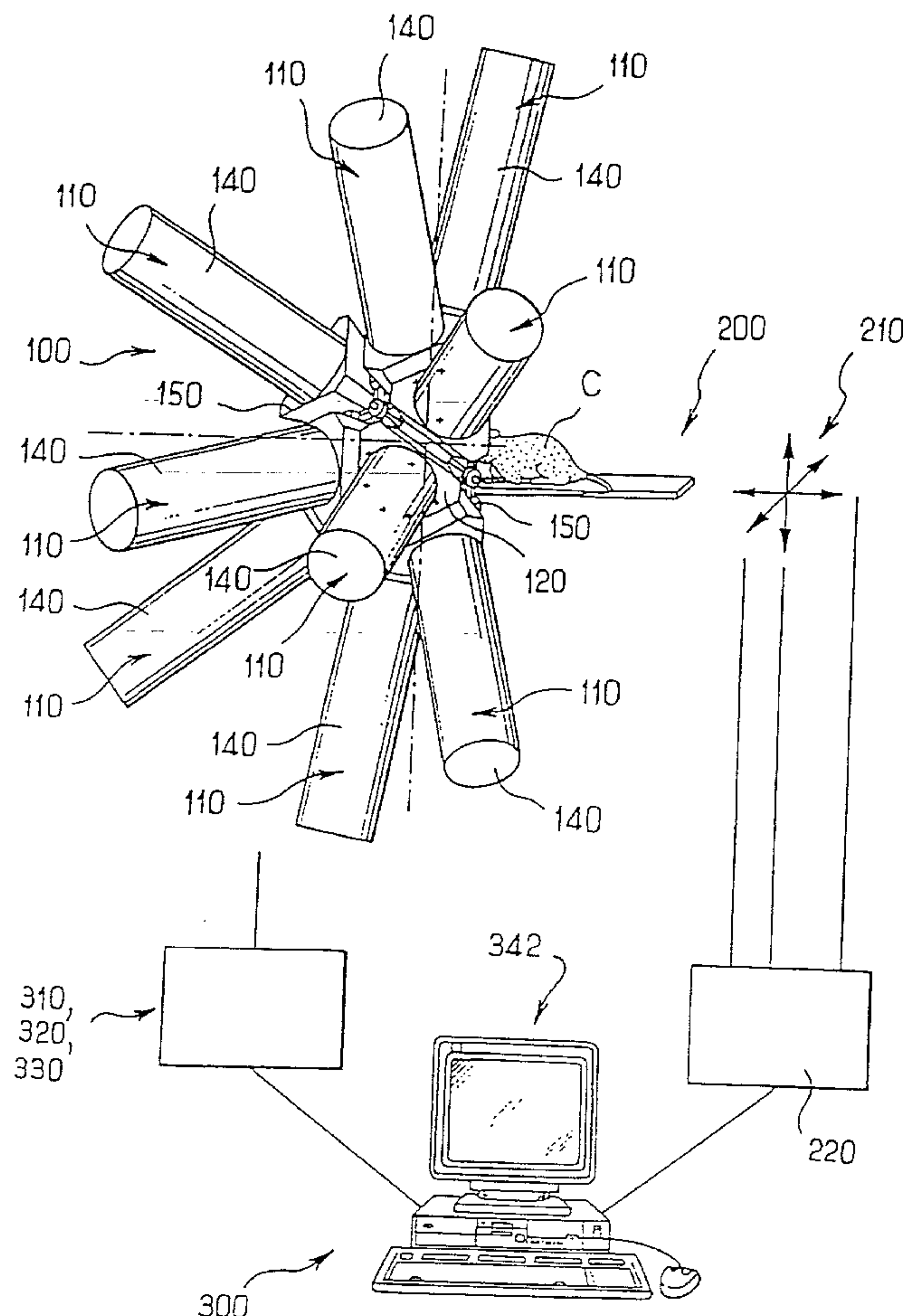
Jun. 20, 1995 (FR)..... 95 07345

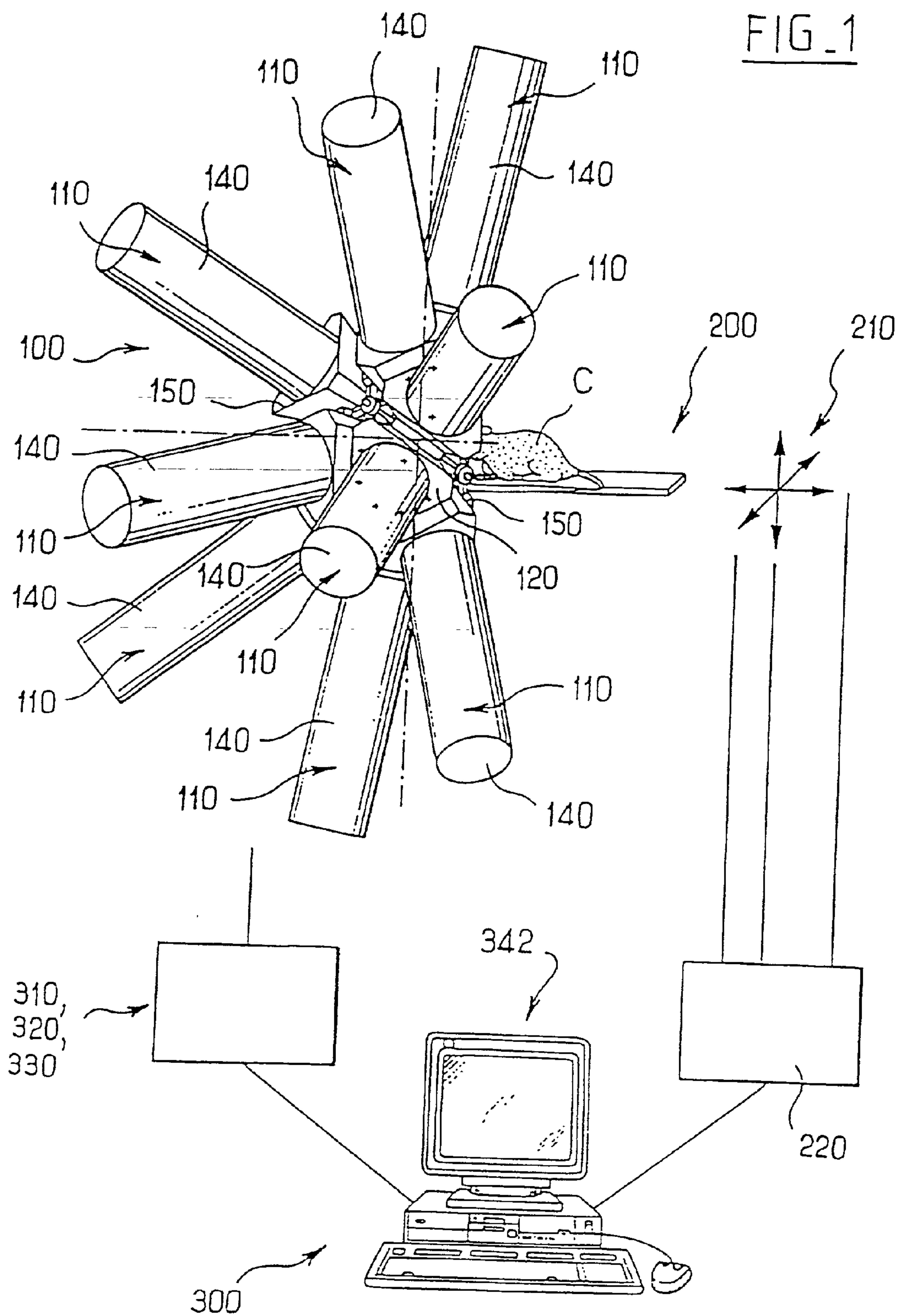
Publication Classification(51) **Int. Cl.⁷** **G21F 1/00**(52) **U.S. Cl.** **250/515.1**

Correspondence Address:

BLAKELY SOKOLOFF TAYLOR & ZAFMAN
12400 WILSHIRE BOULEVARD, SEVENTH
FLOOR
LOS ANGELES, CA 90025 (US)(21) Appl. No.: **09/813,094**(22) Filed: **Mar. 19, 2001**(57) **ABSTRACT**

A non-invasive analysis device including a plurality of sensors (110) combined with collimating structures (120) having a common source focus (O) and processing means (300) providing an AND-type combinational logic function of the output of the sensors (110) for sensing two coincidentally transmitted beams that are at least slightly angularly correlated.





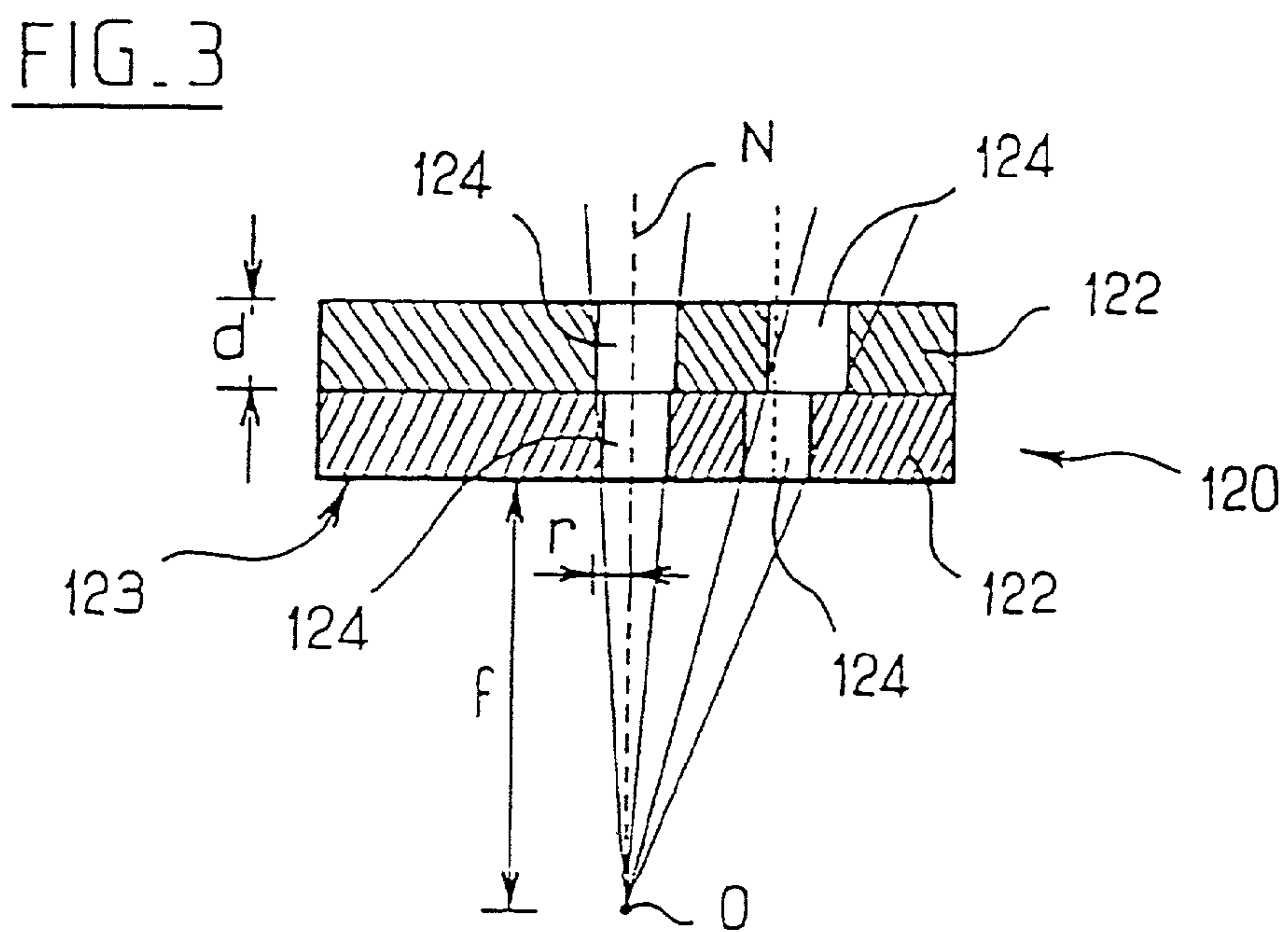
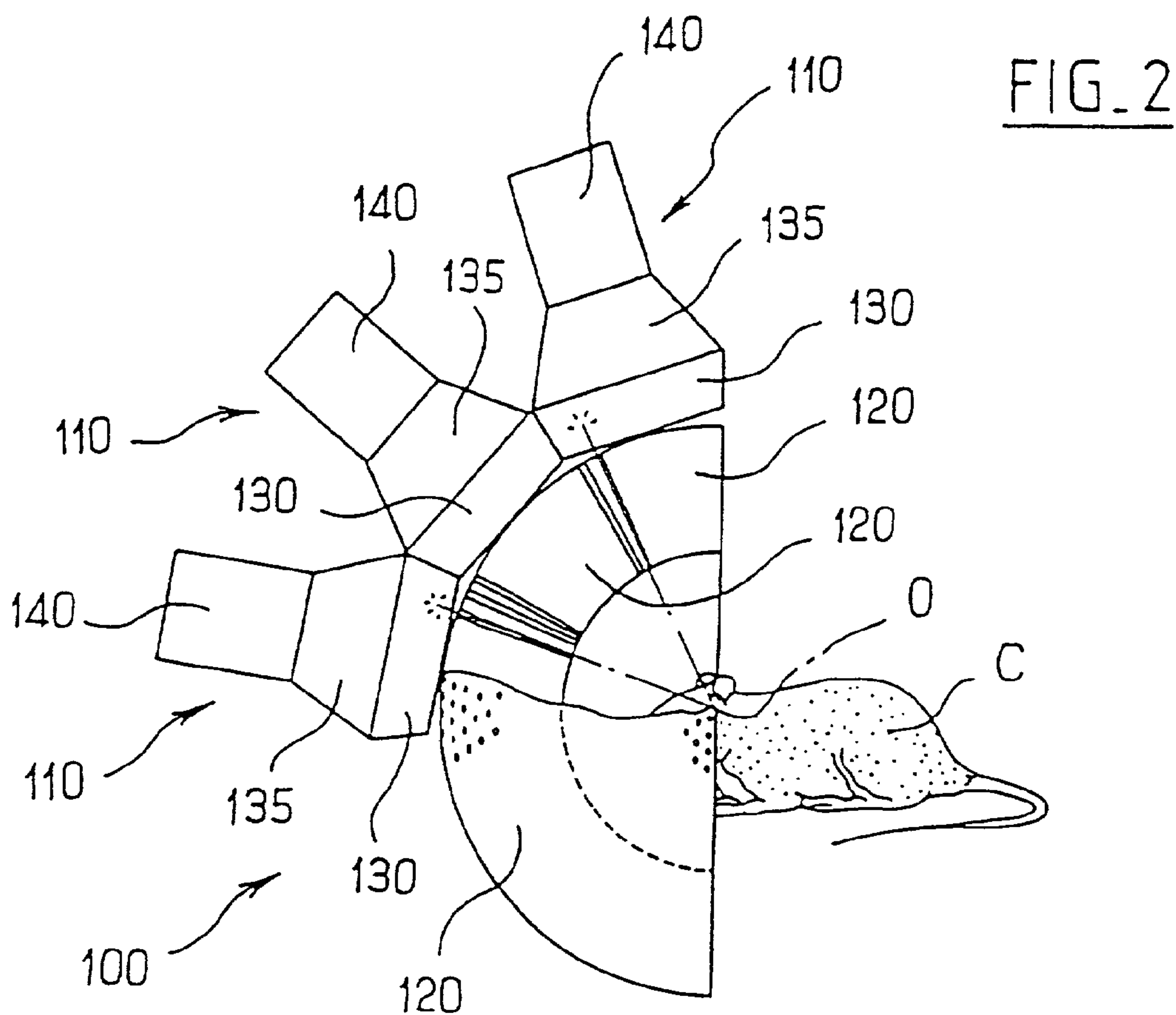


FIG. 4

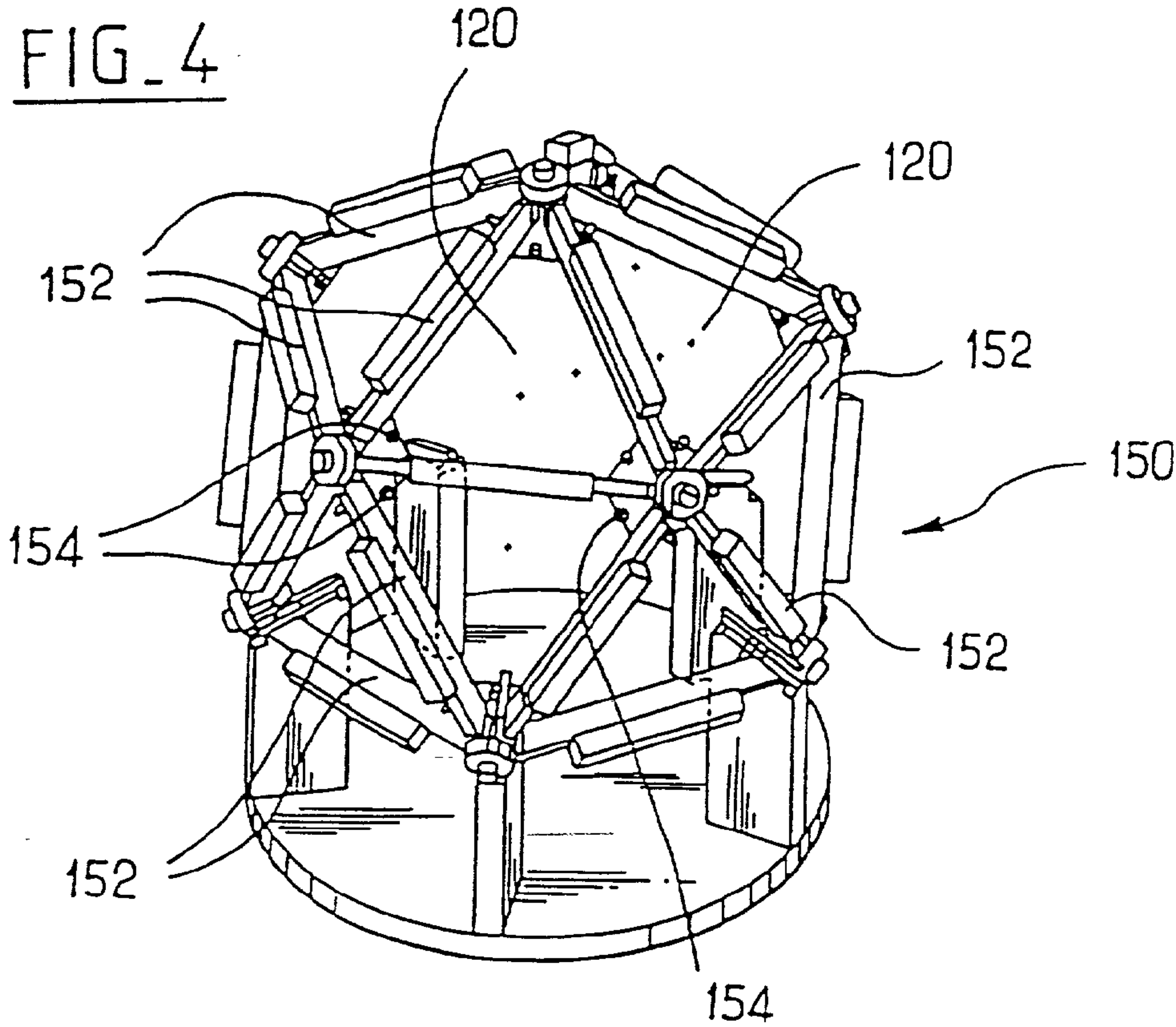
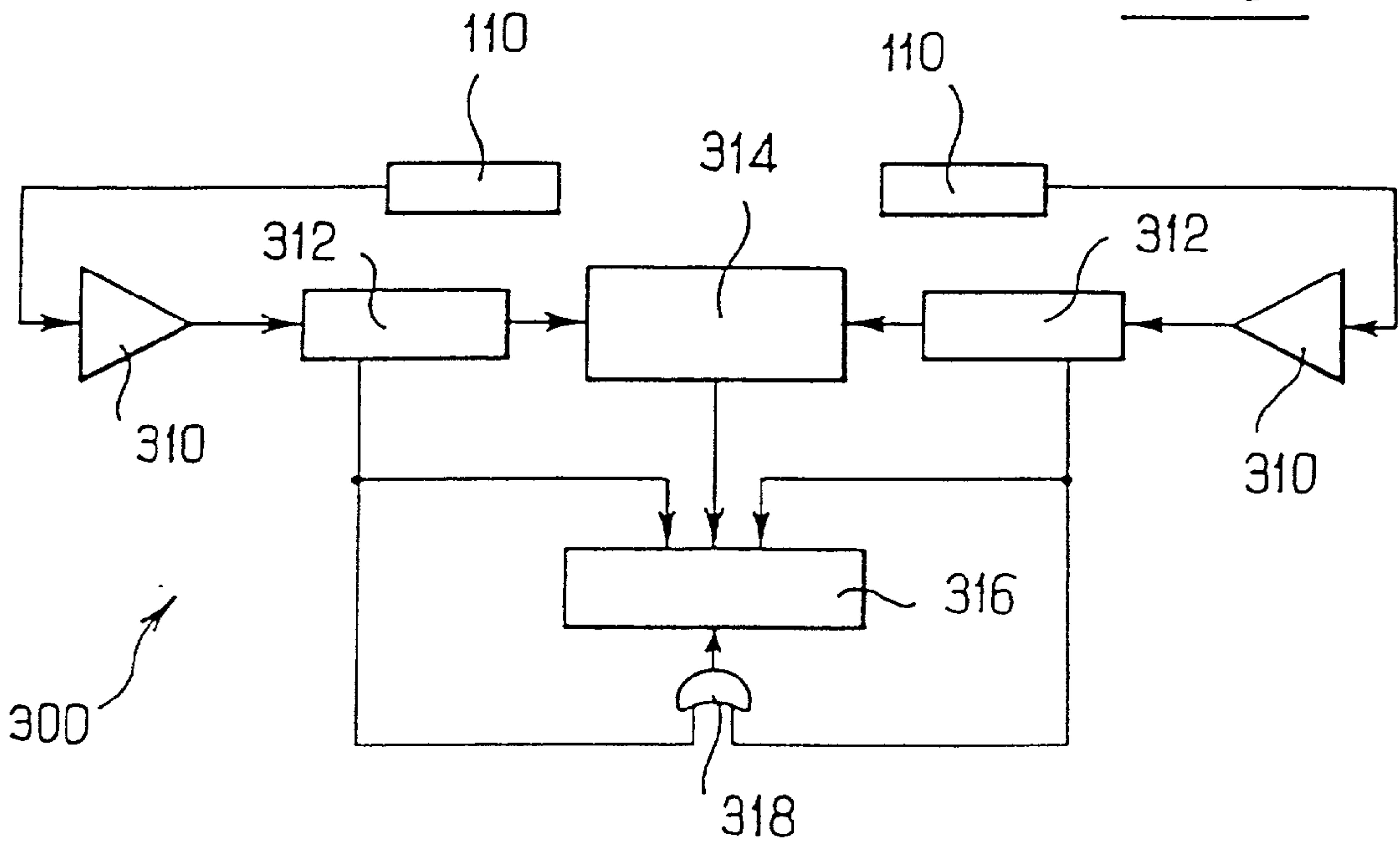
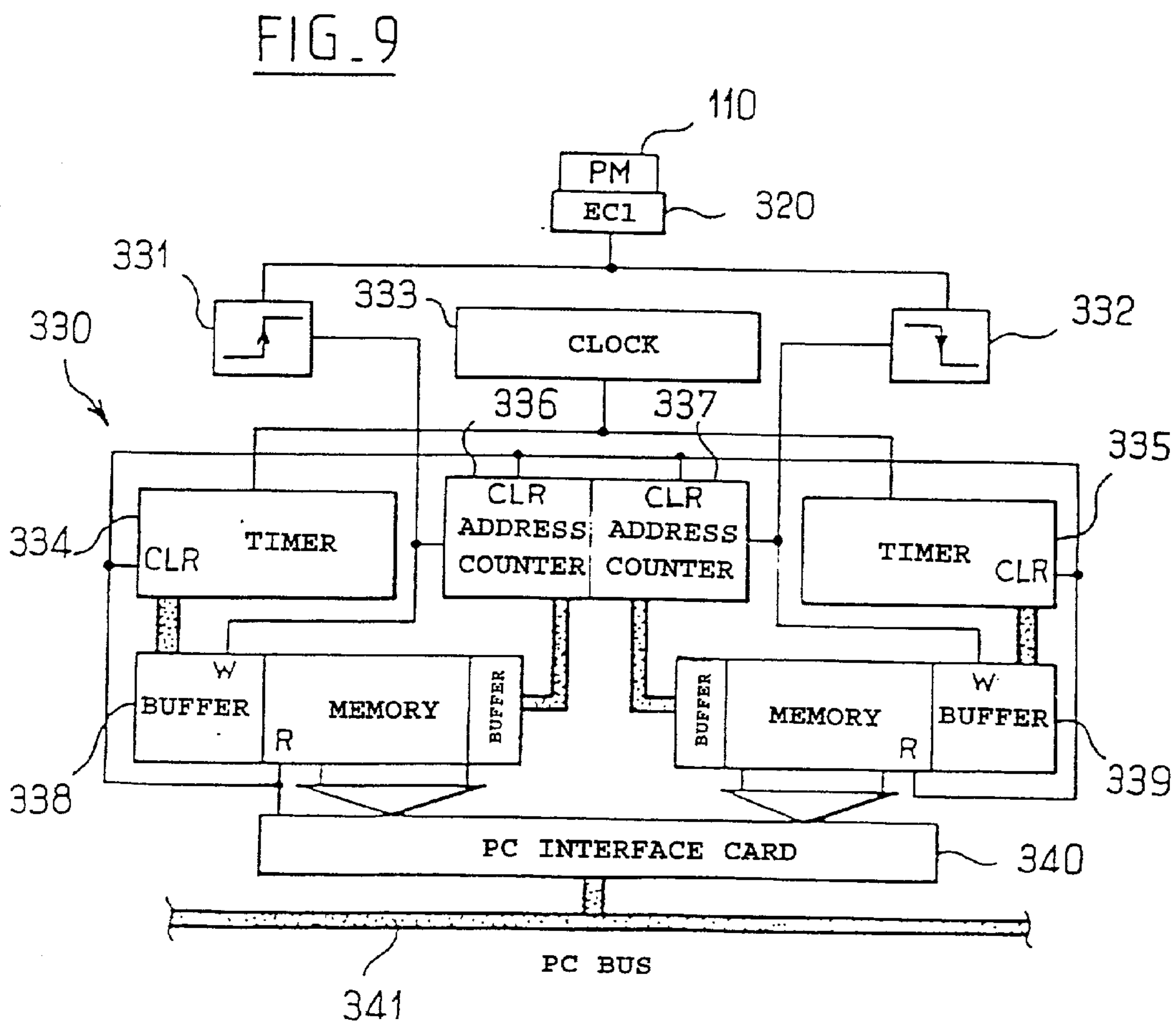
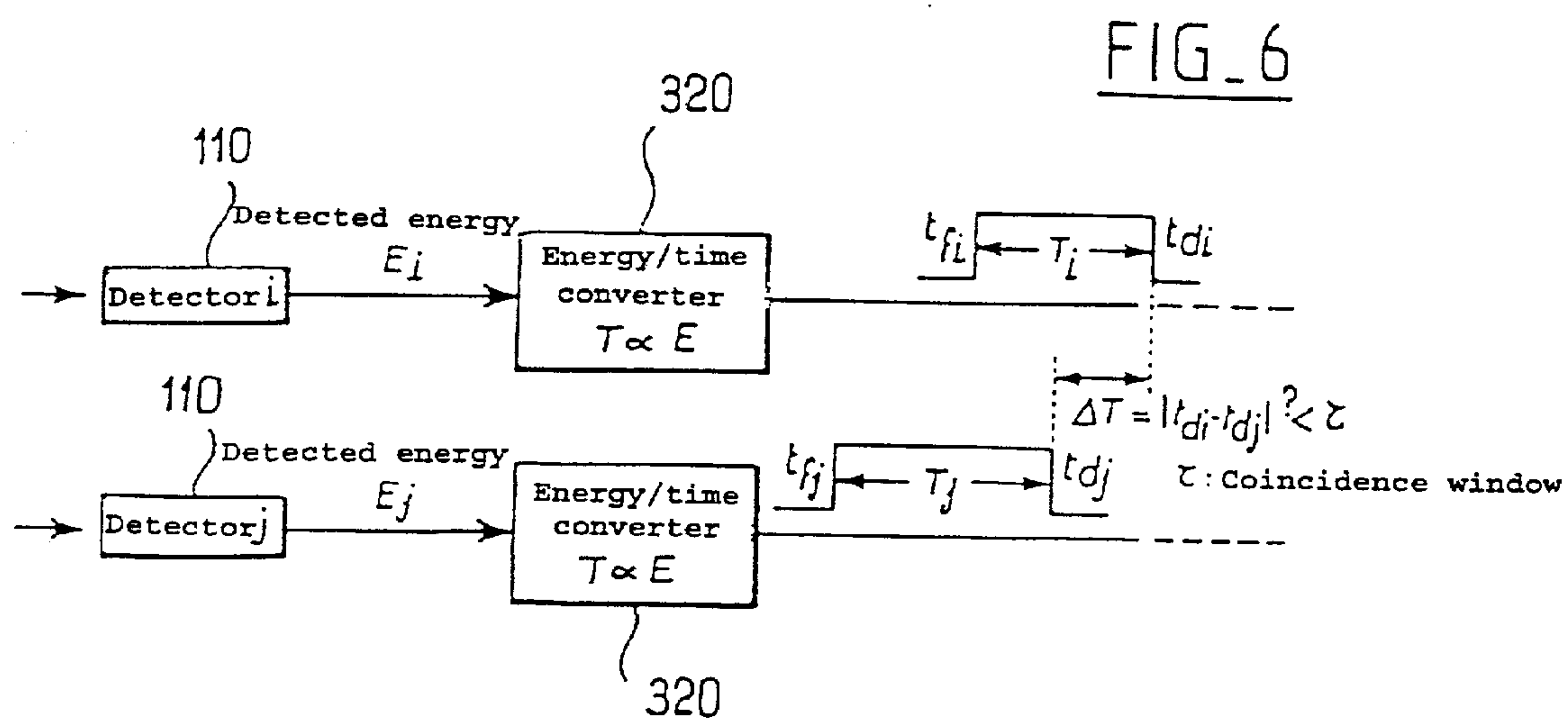


FIG. 5





**DEVICE FOR NON-INVASIVE ANALYSIS BY
RADIO-IMAGING, IN PARTICULAR FOR THE IN
VIVO EXAMINATION OF SMALL ANIMALS, AND
METHOD OF USE**

[0001] The present invention relates to the field of non-invasive devices for analysis by radio-imaging.

[0002] Numerous analysis devices have already been proposed which use radioactive markers (Geiger counters coupled to a collimator, gamma cameras, etc.).

[0003] By utilizing the inherent property of radioactive marking which makes it possible to obtain quantitative information regarding the distribution of the tracer, radio-imaging techniques constitute an important tool both in the clinical field and in the field of fundamental research.

[0004] At present, the devices which are used most widely in this field belong to computer-aided emission tomography.

[0005] Computer-aided tomography has been developed along two different lines: the SPECT (Single Photon Emission Computed Tomography), which uses radio isotopes emitting a single photon by decay, for example ^{99m}Tc , and the PET (Positron Emission Tomography) system, which uses radio isotopes in which two gamma radiation events are emitted simultaneously during the annihilation, in the tissue, of the positron produced by the decay of, for example, ^{18}F .

[0006] Most SPECT systems are based on the use of one or more gamma cameras which are rotated about the object to be analyzed. A typical gamma camera consists of a multi-channel collimator, a large-area scintillator crystal, a light guide for optical coupling between the crystal and a set of photomultiplier tubes, and analog electronics for analyzing the amplitude of the signal and the position encoding. The entire device is contained within lead shielding in order to minimize the background noise produced by sources lying outside the field of view of the camera. The operating principle of a gamma camera is as follows: a photon, produced by a decay event in the source and passing through the collimator, can interact with the scintillator, provoking a local and isotropic scintillation. The photomultiplier tubes located above each receive a light flux which depends on their distance from the light source. It is then possible, on the basis of the electrical signals delivered by each photomultiplier, to reconstruct the position of the scintillation by a center-of-gravity technique and to record and/or send it to a display device.

[0007] Positron emission tomography (PET) is another method which makes it possible to achieve in vivo and non-invasive regional measurement of physiological and metabolic parameters. Positron-emitter radioelements are isotopes having a surplus of protons with respect to their number of neutrons. When a positron is almost at rest, an encounter with an electron gives rise to an annihilation reaction which produces the simultaneous emission of two gamma photons departing in almost opposite directions. PET systems thus comprise an array of detectors in a ring which can detect the coincidence of two photons, as being indicative of the emission of the positron. The site of the annihilation then lies somewhere in the volume defined between the two detectors in question.

[0008] Radio-pharmaceutical imaging constitutes an important tool in the diagnosis, characterization and treat-

ment of diseases and functional disorders. However, before new pharmacological agents are used in man, it is generally necessary to characterize them in animal models in order to determine its biochemical, metabolic and physiological effects.

[0009] Of course, this characterization presupposes the availability of high-resolution imaging techniques in order to evaluate, ex vivo or in vivo, the spatial concentrations of the tracer which is injected.

[0010] At the present time, the spatial resolution of conventional tomographs is from 5 to 7 mm in the case of PET systems and from 8 to 12 mm in the case of SPECT systems. These values prove to be insufficient for carrying out studies in small animals, for example for rat studies of tumors, the typical size of which is of a few mm, or the distribution of neuroreceptors. In actual fact, it is necessary for a tomograph dedicated to the imaging of small animals to be able to provide spatial resolutions of at least ~ 2 mm.

[0011] Since 1990, a number of approaches based on PET and SPECT systems have been pursued in an attempt to achieve the desired performance.

[0012] However, these attempts at improvement have not yet been satisfactory, except at the cost of detection efficiency. The limitations of current tomographs in terms of resolution do not therefore allow in vivo studies to be extended to models on small animals, for which experimentation could be carried out more precisely.

[0013] The object of the present invention is to improve this situation.

[0014] This object is achieved according to the present invention by virtue of an analysis device comprising a plurality of detectors which are associated with collimating structures having a common source focus and processing means performing a combinatorial logic function of the "AND" type on the output of the detectors in order to detect two coincidentally emitted radiation events that are at least slightly angularly correlated.

[0015] According to an advantageous characteristic of the invention, a multi-channel collimator is provided which is formed by a stack of plates having homothetic perforations.

[0016] According to another characteristic of the invention, the perforations in the plates are produced by chemical machining.

[0017] The present invention also relates to an analysis method which comprises the steps consisting in:

[0018] injecting, into a body to be analyzed, a marker which can generate two coincidentally emitted radiation events that are at least slightly angularly correlated, and

[0019] detecting these radiation events using a device of the aforementioned type.

[0020] Other characteristics, objects and advantages of the present invention will emerge when reading the following detailed description, and with reference to the appended drawings which are given by way of non-limiting example, and in which:

[0021] **FIG. 1** represents a general schematic view of an analysis device according to the present invention,

[0022] FIG. 2 represents a view, partially in section, of the detection part of this device,

[0023] FIG. 3 schematically represents a stack of plates forming a collimator,

[0024] FIG. 4 represents a structure for supporting of the detectors,

[0025] FIG. 5 represents a partial diagram of the processing of an acquisition circuit according to a first variant of the invention,

[0026] FIG. 6 schematically represents one form of energy/time conversion of the detected signals, according to a second variant of the invention,

[0027] FIG. 7 represents a circuit diagram for this purpose,

[0028] FIG. 8 represents time diagrams of the signals of this circuit, and

[0029] FIG. 9 represents the general structure of an acquisition circuit.

[0030] The analysis device according to the present invention essentially comprises:

[0031] a set of detectors 100,

[0032] means 200 which are designed to support a body C to be analyzed and to allow a controlled relative displacement between it and the set of detectors 100, and

[0033] processing means 300.

[0034] The set of detectors 100 comprises a plurality of detectors 110 focused on a common source focus O. The detectors 110 are carried by a support structure 150.

[0035] The detectors 110 preferably cover a solid angle at least equal to 2π steradians around the focus O.

[0036] According to the non-limiting particular embodiment represented in the appended figures, fifteen detectors 110 are provided, arranged in correspondence with fifteen adjacent faces of an icosahedron.

[0037] Each detector 110 preferably comprises:

[0038] a collimator 120,

[0039] a scintillator 130,

[0040] an optical guide 135, and

[0041] a photomultiplier 140.

[0042] A detector 110 of this type constitutes a counter of gamma and/or X radiation.

[0043] The collimators 120 are used to select the direction of the photons which are detected. They are formed by collimating structures that focus with large solid angle.

[0044] The focused collimation allows preferential detection of the radiation originating from a small region of space around the focal point O. A collimating structure of this type may be formed by a spherical cap with sectors or a polyhedron consisting of plane parts pierced with conical radial holes and constructed with a material having high photoelectric absorption power.

[0045] The conical radial holes preferably have at least substantially the same entry radii. They also preferably have at least substantially the same exit radii, the same vertex and are juxtaposed in a regular array with axial symmetry. The various channels are separated by partitions whose thickness is tailored to the energy of the radiation emitted by the source, so as to make it possible to absorb a high proportion of those protons whose trajectory is oblique with respect to the channel axis. In this way, only the photons emitted at the focal point will have a significant probability of reaching the scintillators 130.

[0046] In the scope of the present invention, as schematized in FIG. 3, the collimators 120 are preferably made by stacking perforated plates 122 having homothetic perforations 124.

[0047] The plates 122 are advantageously made of tungsten.

[0048] The reason for choosing tungsten is, on the one hand, its high absorption power: it has an absorption coefficient which is 30% to 40% greater than that of lead in the 10-500 keV range. Further, its mechanical properties ensure that the system is rigid and that the shape of the holes 124 is precise.

[0049] Moreover, within the scope of the invention, the homothetic perforations 124 made in the tungsten plates 122 are preferably obtained by chemical machining.

[0050] After lengthy studies and experiments, this technique was found to be superior to other known piercing techniques, such as laser piercing or wire spark erosion.

[0051] Machining by chemical etching consists in depositing a photosensitive resin on all the parts which are to be preserved, by making use of a mask representing the part to be produced. The part is then immersed in a bath which chemically etches the unprotected regions to form the desired openings 124.

[0052] As schematized in FIG. 3, the holes 124 in each plate 122 are cylinders that are parallel to one another and orthogonal to the faces of the plates 122. The result of this is that the solid angle, with respect to the source focus O, subtended by the aperture of a hole 124, decreases when moving away from the normal N to the plates 122 which passes through their homothetic center O.

[0053] The radius r of the holes 124 in the entry face 123, the thickness d of the plates 122 and the focal length f are determined so as to conserve an acceptable solid angle subtended by the aperture of each hole 124 with respect to O.

[0054] The thickness d of the plates 122 is preferably less than the diameter of the holes 124 in the internal entry face 123 of the collimator, for example equal to half this diameter, and the thickness of the span (septum) between the holes 124, in this internal face 123, is greater than or equal to the thickness of the plates 122 (i.e., for example, a distance between the centers of the holes to equal to three times the radius of the holes 124 on the internal face 123).

[0055] If the detectors 110 are assembled on a polyhedral support 150 of the regular icosahedron type, each collimation module 120 takes the form of a truncated triangular pyramid.

[0056] According to one non-limiting particular embodiment, each collimator **110** is formed by stacking 48 tungsten plates **122** having a thickness of 0.2 mm and an array of holes **124** defining a focal length f of the order of 7 cm, the holes **124** having, in the internal face **123** of the collimator, a radius of the order of 0.2 mm and an interaxial distance of the order 0.5 mm.

[0057] More precisely, each collimator **110** may be formed by stacking 24 different pairs of pairwise identical plates **122**. In an arrangement of this type, each plate, of thickness d , is formed by juxtaposing two identical screens of thickness $d/2$. In this way, it is easy to produce holes **124** of diameter equal to d , with a separation between centers equal to $3d/2$.

[0058] Yet more precisely, according to a non-limiting very particular embodiment, the radius of the holes **124** is 0.205 mm in the first plate **122** and 0.231 mm in the last plate **122**, the distance between centers of the holes **124** is 0.614 mm in the first plate **122** and 0.693 mm in the last plate **122**, the length of the edges of the first plate **122** is 87.2 mm and 98.3 mm for the last plate **122**, and the distance to the focal point is 71.6 mm for the first plate **122** and 80.8 mm for the last plate **122**.

[0059] The scintillator **130** is preferably formed by a single crystal of thallium-activated sodium iodide (NaI(Tl)).

[0060] In the case of an arrangement of the icosahedral type, the scintillator crystal has the form of a prism with triangular cross section in order to cover the entire exit surface of the collimator.

[0061] The light guide **135** provides optical coupling between each scintillator crystal **130** and the associated photomultiplier tube **140**.

[0062] The structure **150** which supports the detectors **110** has to position the focal points of the collimators **120** with sufficient precision, typically of the order of 0.1 mm. There are a number of possible alternative embodiments for a support structure **150** of this type.

[0063] One particular embodiment, comprising a framework consisting of support bars **152** connected in sets of 5 by their ends and arranged along the edges of an icosahedron, is represented in **FIG. 4**. The plates **122** of the collimators **120** can be fixed on these bars **152** using rods **154** engaged in the corners of the plates **122**.

[0064] However, as a variant, the structure **150** may support the light guides **135** of the detectors, instead of the collimators **120**.

[0065] The system **200** for supporting and displacing the analyzed object is designed to define three degrees of freedom in linear displacement for the analyzed object **C** with respect to the sensors **110**. These displacements can be brought about using three controlled and identified motorized shafts, schematically represented at **120** in **FIG. 1**, which are mutually orthogonal and associated with control means **220** which displace the motors interactively in order to position the object **C** and define an analysis region and a displacement step in order to automatically carry out the scan needed for acquisition.

[0066] The mechanism **210** for displacing each shaft may be composed of the following units for each motorized shaft:

[0067] a translation table which drives the movement of a carriage by a screw/nut system, over a travel of, for example, 10 cm, and

[0068] a stepper motor which allows controlled increments of, for example, 10 μm on each translation table.

[0069] The image resulting from an acquisition is constructed on the basis of the number of photons detected at each position of the displacement system **210**. During the acquisition, it is therefore necessary to avoid any movement of the analyzed object **C** with respect to the displacement system **210**, so as not to introduce artefacts in the image.

[0070] The system **200/210** consequently needs to be equipped with a system for specific positioning of the analyzed object.

[0071] By way of non-limiting example, in the case of imaging the brain of a rat, it is simply sufficient to immobilize the rat's head. The positioning system used in this case may correspond to those which are known in stereotaxy equipment. In this case, the head is immobilized at three points: at the entries of the auricular orifices, with two bars of adjustable position, and behind the incisors, with a bar on which the upper jaw bears.

[0072] The processing means **300** are designed to detect two temporally coincident radiation events produced on all of the detectors **110**, without possibly taking into consideration coincidences produced in the same detector or those produced in diametrically opposite detectors.

[0073] An event is thus detected with all the more probability that it originates from the focal point by virtue of the combination of physical collimation, given by the collimating structure, and electronic collimation, given by the coincident detection of the photons produced by the decay.

[0074] When two photons of equal energy are to be detected, the processing and acquisition electronics **300** may be particularly simple, as represented in **FIG. 5**.

[0075] Each detection line associated with a detector **110** comprises an amplifier **310** and a discriminator module **312** which define a window around the photopeak. The outputs of the single-channel analyzers **312** are used to generate the coincidence signal in a time/amplitude converter **314**. Three signals are sent to an interface card **316**: the coincidence signal output by the converter **314** and the two outputs of the single-channel analyzers **312**, in the latter case with the object of constructing a single-photon image for each detector **110**. An acquisition signal, generated by logic combination in a gate **318** of the single-event images originating from the discriminators **312**, is also applied to the interface card **316**.

[0076] The device which has been described above and is illustrated in **FIG. 5** nevertheless soon becomes bulky and impractical when the intention is for coincident detection, and from more than two detectors **110**, of two radiation events of different energy.

[0077] In order to resolve this difficulty, a specific acquisition device is proposed in the scope of the invention, which makes it possible, for each detected event, to extract the energy of the event and the time at which it occurs, in order to test its temporal coincidence with another event.

[0078] A device of this type carries out, for example, energy/time conversion and encodes the events in a signal whose width is proportional to their energy. A system of this type also permits intrinsic encoding of the time of arrival of each event, and therefore permits coincidence tests.

[0079] The operating principle of a system for detecting coincidences which is based on an energy/time converter is as follows. Two events are detected on the detectors *i* and *j* respectively. Each energy/time converter produces a signal whose width *T* is proportional to the energy imparted to the detector. If the widths *T_i* and *T_j* correspond to the energies involved, the events are considered to be coincident if the difference between the start of the signal, $t_{di}-t_{dj}$, is less than the width of the coincidence window *T* fixed beforehand.

[0080] FIG. 7 represents an illustrative embodiment of a circuit 320 according to the invention, which can perform energy/time conversion of this type, and FIG. 8 represents time diagrams of the signals taken from various points in the circuit.

[0081] The circuit in FIG. 7 comprises:

[0082] a capacitor 321 capable of integrating the signal output by a photomultiplier 140,

[0083] an output signal detector 322 for the photomultiplier,

[0084] a delay cell 323 initiated by the aforementioned detector 322,

[0085] a current source 324 which is connected in parallel with the capacitor 321 and is driven by the output of the delay cell 323,

[0086] a comparator 325, one input of which is grounded and the other input of which is connected to the capacitor 321, the output of this comparator 325 constituting the output of the energy/time conversion device, and

[0087] a switch 326 which is controlled by the comparator 325 and can discharge the capacitor 321.

[0088] This circuit 320 operates as follows.

[0089] The scintillation light produced by the interaction of an X or a γ photon in the NaI(Tl) crystal 130 is manifested, on the anode of the photomultiplier 140, as a signal having a very fast rise (in absolute value) followed by a virtually exponential fall, typically with a time constant of the order of 230 ns. Since the integral of this response is proportional to the energy imparted by the radiation to the crystal, the object of the energy/time converter 320 is to recover this integral in order to modulate the width of a square-wave signal. The integral of the anode signal is recovered on the capacitor 321 over a predefined time. This time is obtained on the basis of the delay applied to the signal extracted from the last dynode of the photomultiplier 140 (signal detection 322 and delay cell 323 modules). At the end of the integration time, the current generator 324, which is connected in parallel with the capacitor 321, is turned on in order to bring about a linear discharge of this capacitor. At the same time as the current generator 324 is started, the output of the flip-flop 325 which gives the output signal of the converter 320 changes to a logic state "1". Wherein the voltage of the capacitor 321 passes through zero, the flip-flop 325 returns to its low state, thus interrupt-

ing the high level of the output signal, the current generator 324 is stopped and the capacitor 321 is discharged fully by operating the switch

[0090] FIG. 9 illustrates an illustrative embodiment of an acquisition circuit 330 according to the embodiment.

[0091] This circuit 330 is designed to encode energies which, for example, range from ~10 keV (27 keV for ^{123}I and ^{125}I) to ~300 keV (245 keV for ^{111}In) at a maximum counting rate of $\sim 10^4$ hits per second per detector.

[0092] The system 330 is, for example, composed of:

[0093] a set of (for example 15) detectors (scintillator 130+photomultiplier 140),

[0094] one energy/time converter 320 coupled to each detector 110,

[0095] two detectors 331, 332, respectively for leading and trailing edges of the signals output by the converters 320,

[0096] a clock 333 for the time base of the signals,

[0097] two timers 334, 335 driven by the clock 333,

[0098] two address counters 336, 337 driven by the edge detectors 331, 332, and

[0099] two memories 338, 339 for cyclically and temporarily storing information then transferring it on to the bus 341 of a computer 342.

[0100] At the output of each energy/time converter 320, a signal is found whose width is proportional to the energy imparted to the detector 110. This width may, for example, vary from ~10 ns for 3 keV to ~1 μs for 300 keV. This readily permits a counting rate of 10^5 hits per second. The times of the leading and trailing edges of the signal are calculated using the timers 334, 335 and clock 333, and are stored in the independent memories 338, 339. Once acquisition has been completed in a voxel, the data in the memories 338, 339 are transferred to the bus 341 of the PC in order to be processed using specific software.

[0101] Of course, the data acquisition method according to the present invention, with a view to imaging, comprises the initial step of injecting the body C to be analyzed with a radioactive marker capable of emitting two coincident radiation events that are at least slightly angularly correlated.

[0102] There are a number of possible variants for a marker of this type.

[0103] Attention may be drawn to at least three mechanisms of radioactive de-excitation which give rise to the emission of two coincident photons that are only slightly angularly correlated.

[0104] In the first of the de-excitation mechanisms (isobaric de-excitation of the nucleus), the decay of a parent nucleus produces an excited nucleus which changes to its ground state through a cascade, thus producing two coincident gammas. An example of a radioelement which de-excites by this mechanism is ^{111}In . After an electron capture, ^{111}In (half-life: 2.8 days), changes essentially to the 417 keV excited level of ^{111}Cd . The ground state of ^{111}Cd is reached by the cascade emission of one gamma of 171 keV followed by another with 245 keV; the half-life of the 245 keV level is 85 ns.

[0105] The other two mechanisms relate to radioelements whose decay starts with electron capture, and this serves to provide the first photon of the coincident pair. Specifically, in electron capture, the X_k emission resulting from the rearrangement of the inner electron shells is utilized to provide a first photon. The de-excitation of the daughter nucleus gives rise to the second photon, and there are two possible ways that this may take place. If the de-excitation is radiative, there will be a γ photon coincidence with the X_k photon. If the de-excitation takes place through internal conversion, there will also be electron rearrangement and emission of an X_k photon in coincidence with that of the electron capture. Two isotopes of iodine ($Z=53$) give an example of these two coincidence mechanisms: ^{123}I (half-life: 13 h) and ^{125}I (half-life: 60 days). In both cases, the emitted X_k radiation has an energy of about 27 keV, the probability of capturing an electron in the K shell is 80% and the fluorescence efficiency is 86%. In the case of ^{123}I , the electron capture leads to an excited state of ^{123}Te , which changes to the ground state by emitting a 159 keV gamma, the X_k - γ coincidence factor is about 70%. In the case of ^{125}I , the excited state of the daughter nucleus at an energy of 35 keV, and the change to the ground state takes place either by emission of a gamma (7% probability) or by internal conversion; the coincidence factor for $X_k\gamma$ plus X_k - X_k is 60%.

[0106] The invention is not limited to the use of the aforementioned radioelements: ^{123}I , ^{125}I and ^{111}In , but extends to any other equivalent radioelement which can emit at least two temporarily coincident radiation events, for example γ - γ , γ - X or X - X , that are at least slightly angularly correlated.

[0107] After or before the radio tracer is injected, the body C to be analyzed is placed in the collimating detection structure 100.

[0108] The radiation events which are emitted by the object C and pass through the collimators 120 are detected by all the detectors 110 (scintillators 130+photomultipliers 140). The signal which is output by the detectors 110 and has been amplified and processed in the means 310, 320, 330 is sent to a computer 342 whose task is to acquire and store it. The processing electronics 310, 320, 330 deliver either the signals indicating coincidence between any two detectors 110, or a set of signals so that the coincidences which occur can be recovered a posteriori.

[0109] The coincidence detection makes it possible to optimize the spatial resolution and to reject signals originating from points lying outside the focal point.

[0110] Displacing the focal point of the collimators 120 by scanning through the volume of the region of interest during the acquisition makes it possible to construct the image of the object C voxel by voxel. It is thus unnecessary to use an algorithm for reconstructing the image, which may in particular amplify the statistical fluctuations on the reconstructed image, as in the case of the SPECT and PET techniques, and the image may, according to the invention, be displayed voxel by voxel as the acquisition proceeds.

[0111] The present invention may give rise to a number of applications which require measurement of the concentration of radioactively marked chemical species, non-limiting examples which may be mentioned of which applications include the in vivo examination of small animals, in par-

ticular within the scope of clinical research, relating for example to the detection of cardiovascular lesions, oncology, the detection and monitoring of tumors, studying the distribution of neuroreceptors, displaying the functions of the brain (in the case of neurodegenerative diseases, such as Parkinson's or Alzheimer's diseases, or Huntington's disease; or in the event of psychiatric disorders, such as in schizophrenia), gene therapy or, more generally, neurobiology or neuropharmacology in order to evaluate the effectiveness of treatments based on the administration of neuroprotective agents and on neural grafting.

[0112] The present invention is not, of course, limited to the particular arrangements which have just been described, but extends to any variant in accordance with its spirit.

[0113] The use of the device and the implementation of the method which were described above can be carried out by any authorized individual without requiring particular knowledge in the medical field.

1. Device for non-invasive analysis, characterized in that it comprises a plurality of detectors (110) which are associated with collimating structures (120) having a common source focus (O) and processing means (300) performing a combinatorial logic function of the "AND" type on the output of the detectors (110) in order to detect two coincidentally emitted radiation events that are at least slightly angularly correlated.

2. Device according to claim 1, characterized in that it comprises means for counting radioactive radiation events.

3. Device according to one of claims 1 or 2, characterized in that it further comprises means (200) which are designed to support the body (C) to be analyzed, and to allow a controlled relative displacement between it and the set of detectors (110).

4. Device according to one of claims 1 to 3, characterized in that the detectors (110) cover a spatial field at least equal to 2π steradians around the focus (O).

5. Device according to one of claims 1 to 4, characterized in that the detectors (110) are arranged in accordance with the faces of an icosahedron.

6. Device according to one of claims 1 to 5, characterized in that it comprises fifteen detectors (110).

7. Device according to one of claims 1 to 6, characterized in that each detector (110) comprises a collimator (120), a scintillator (130), an optical guide (135) and a photomultiplier (140).

8. Device according to one of claims 1 to 7, characterized in that each collimating structure (120) comprises conical radial holes (124).

9. Device according to one of claims 1 to 8, characterized in that it comprises a structure (150) which is intended to support the set of detectors (110) and is in the form of a framework consisting of bars (152) connected by their ends.

10. Device according to one of claims 1 to 9, characterized in that it comprises means (200) which can effect a relative displacement which is controlled and identified on the basis of three orthogonal axes between the body (C) to be analyzed and the set of detectors (110).

11. Device according to one of claims 1 to 10, characterized in that it comprises processing means (300) comprising energy/time conversion means (320) which can encode the detected events in the form of signals whose width is proportional to the energy of the event.

12. Device according to claim 11, characterized in that the processing means (320) comprise:

- a capacitor (321) capable of integrating the signal output by a photomultiplier (140),
- an output signal detector (322) for the photomultiplier,
- a delay cell (323) initiated by the aforementioned detector (322),
- a current source (324) which is connected in parallel with the capacitor (321) and is driven by the output of the delay cell (323),
- a comparator (325), one input of which is grounded and the other input of which is connected to the capacitor (321), the output of this comparator (325) constituting the output of the energy/time conversion device, and
- a switch (326) which is controlled by the comparator (325) and can discharge the capacitor (321).

13. Device according to one of claims 11 and 12, characterized in that the processing means (300) comprising timers (334, 335) which are driven by a clock (333) and are designed to define the trip times of the signal output by the energy/time conversion means (320).

14. Device according to claim 13, characterized in that the timers (334, 335) are controlled by detectors (331, 332) of leading and trailing edges of the signals output by the energy/time convertors (320).

15. Collimator for a radio-imaging analysis device, characterized in that it is formed by a stack of plates (122) provided with perforations (124).

16. Collimator according to claim 15, characterized in that the perforations (124) provided in the various plates (122) exhibit a homothetic progression.

17. Collimator according to one of claims 15 and 16, characterized in that the perforations (124) are produced by chemical machining.

18. Collimator according to one of claims 15 to 17, characterized in that the plates (122) are made of tungsten.

19. Collimator according to one of claims 15 to 18, characterized in that the thickness (d) of the plates (122) is less than the diameter of the perforations (124) in the internal entry face (123) of the collimator and the span between the perforations (124) is greater than the thickness of the plates (12).

20. Collimator according to one of claims 15 to 19, characterized in that at least some of the plates (122) are formed by superposition of two identical perforated screens.

21. Analysis method, characterized in that it comprises the steps consisting in:

injecting, into a body (C) to be analyzed, a marker which can generate two coincidentally emitted radiation events that are at least slightly angularly correlated, and

detecting these radiation events using an analysis device comprising a plurality of detectors (110) which are associated with collimating structures (120) having a common source focus (O) and processing means (300) performing a combinatorial logic function of the "AND" type on the output of the detectors (110) in order to detect two coincidentally emitted radiation events that are at least slightly angularly correlated.

22. Method according to claim 21, characterized in that the marker which is used is designed to emit two coincidentally emitted radiation events that are at least slightly angularly correlated through atomic (X emitted following electron capture or internal conversion) or nuclear origin (gamma emitted during the isobaric de-excitation of a nucleus).

23. Method according to one of claims 21 and 22, characterized in that the marker which is used is chosen from the group comprising ^{123}I , ^{125}I , ^{111}In .

24. Method according to one of claims 21 to 23, characterized in that the image of the body to be analyzed is constructed voxel by voxel.

* * * * *