

[54] **METHOD FOR DETERMINING DISCRETE LOCALIZED BLOOD FLOW IN A LARGE VOLUME OF TISSUE**[75] Inventors: **Myron Youdin**, Flushing; **June N. Barker**; **Theobald Reich**, both of New York, all of N.Y.[73] Assignee: **New York University**, New York, N.Y.[22] Filed: **Apr. 26, 1973**[21] Appl. No.: **354,489****Related U.S. Application Data**

[63] Continuation-in-part of Ser. No. 126,770, March 22, 1971, Pat. No. 3,769,966.

[52] U.S. Cl. **128/2.05 F; 128/2 A; 250/303**[51] Int. Cl. **A61b 5/02; A61b 6/00**[58] Field of Search **128/2.05 F, 2.05 R, 2.05 A, 128/2.05 V, 2 A, 2 G, 2 R; 250/303, 363, 364**[56] **References Cited****UNITED STATES PATENTS**

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Primary Examiner—**Kyle L. Howell**Attorney, Agent, or Firm—**Hubbell, Cohen and Stiefel**

[57]

ABSTRACT

A method for determining discrete localized blood flow in a large volume of tissue, such as local cerebral emission flow in a discrete radiation emission detection zone of a patient's brain. The brain is subdivided into a plurality of detection zones, such as 144, by means of a headgear apparatus which defines these detection zones by means of a multiplanar array of an equal number of detectors, each detector having an associated collimator to limit the field of view of the detector. A radioactive gas, such as one having two gamma energy levels, such as Xe¹³³, or a gaseous mixture of two different gamma energy level isotopes, is inhaled by the patient, such as in a closed inhalation-exhalation breathing system, over a predetermined data collection period during which the concentration of the gas in the blood is periodically increased. Determinations of arterial blood radiation concentration and tissue radiation concentration are made during this period, and the local cerebral blood flow is subsequently determined from these determinations. The arterial blood radiation concentration is preferably determined by determining the radiation concentration of the gas in the last part of an exhalation, while the tissue radiation concentration is preferably determined from the radiation emissions quantified by the detector array such as by detecting both gamma energy levels, the field of view of the detector depending on the gamma energy level being detected.

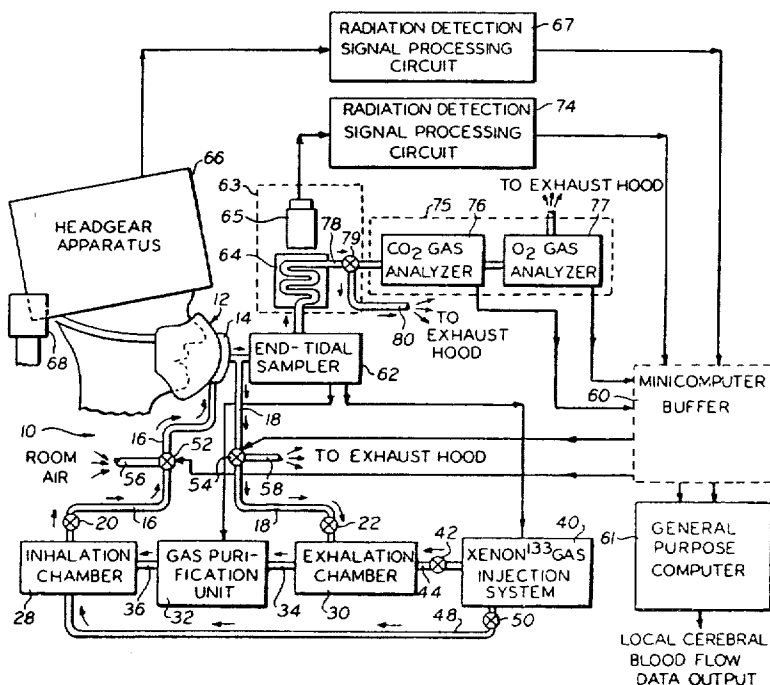
59 Claims, 59 Drawing Figures

FIG. 1.

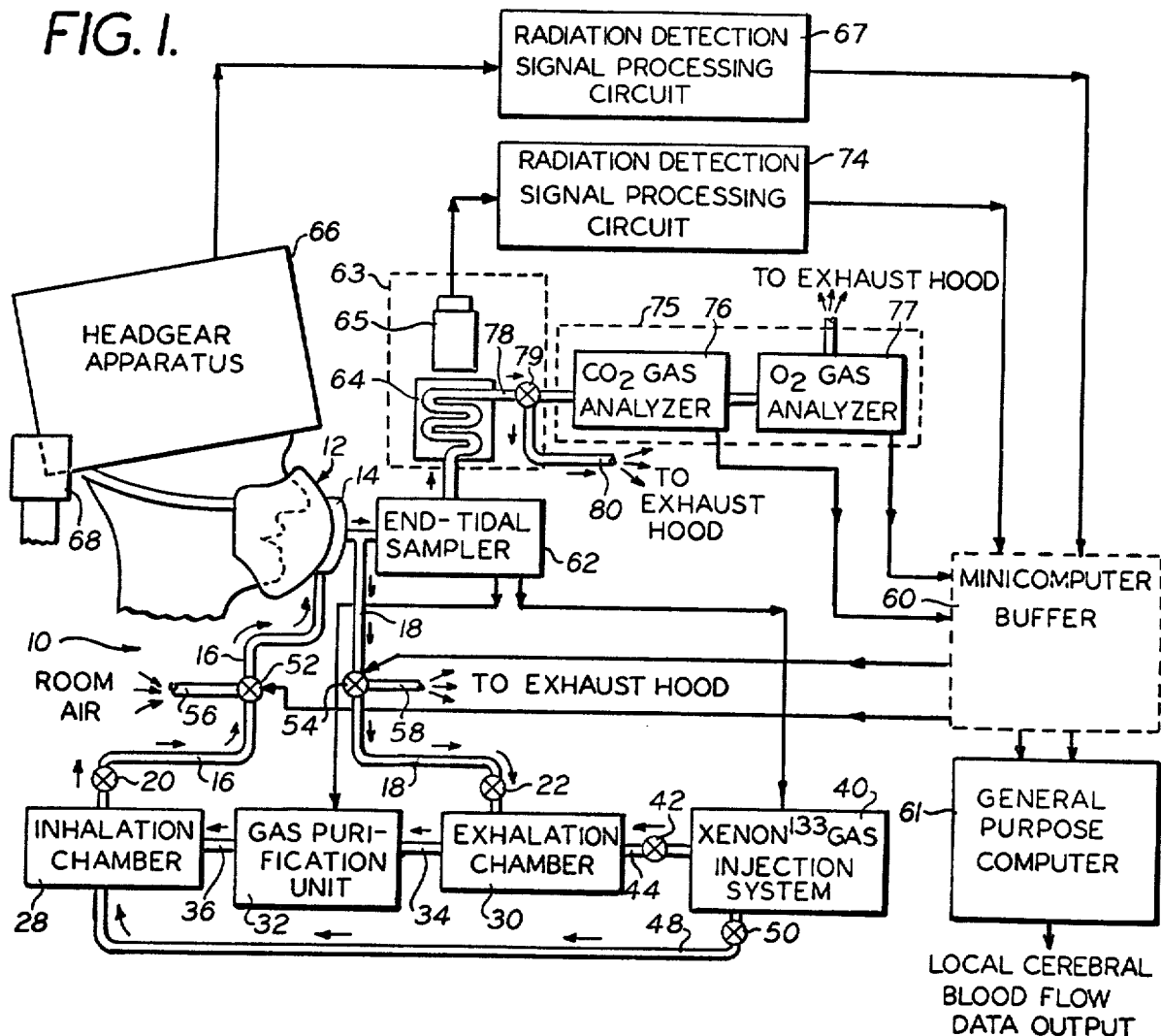
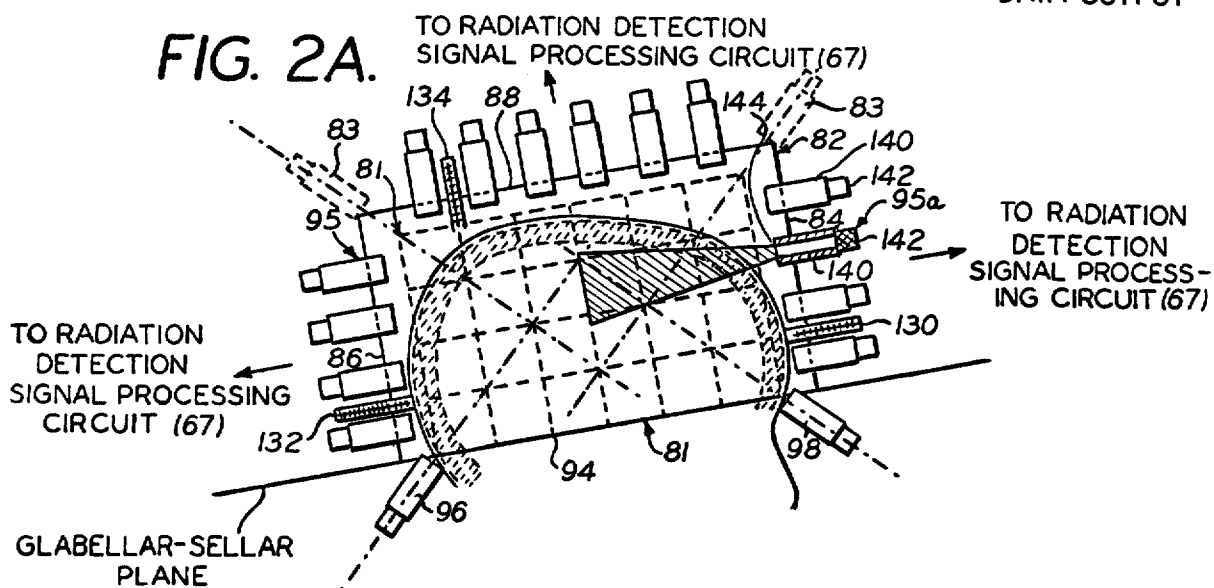


FIG. 2A.



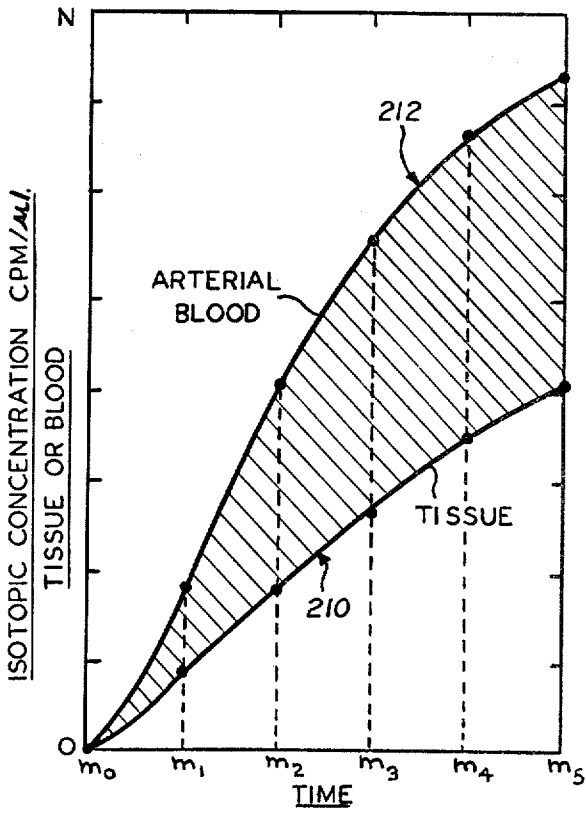
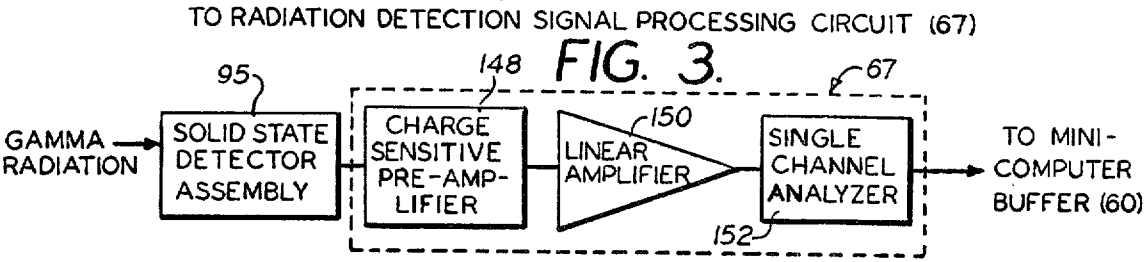
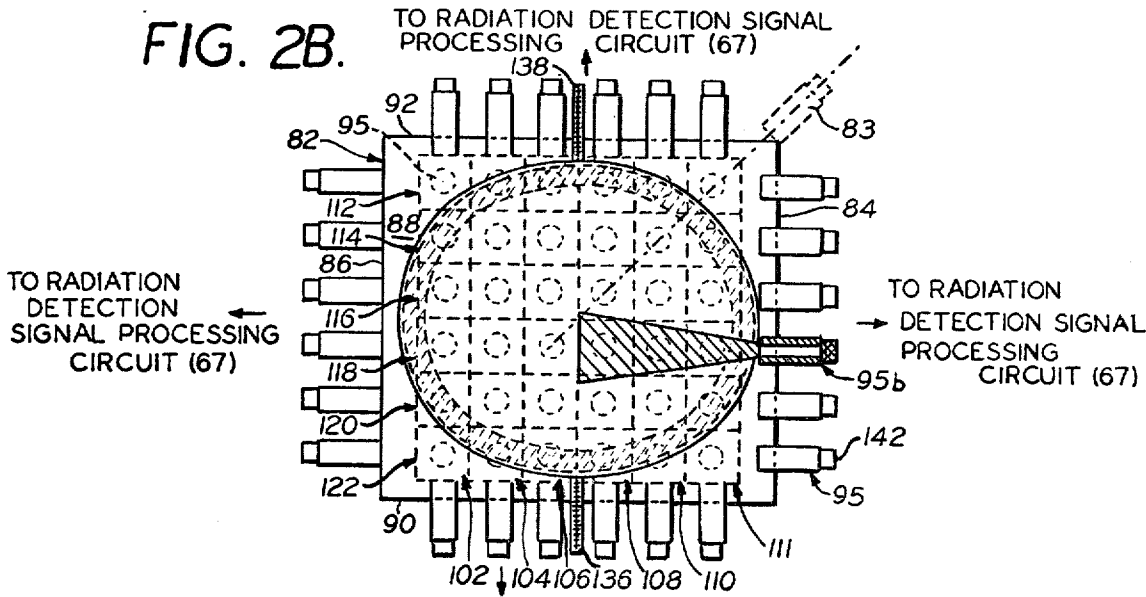


FIG. 4.

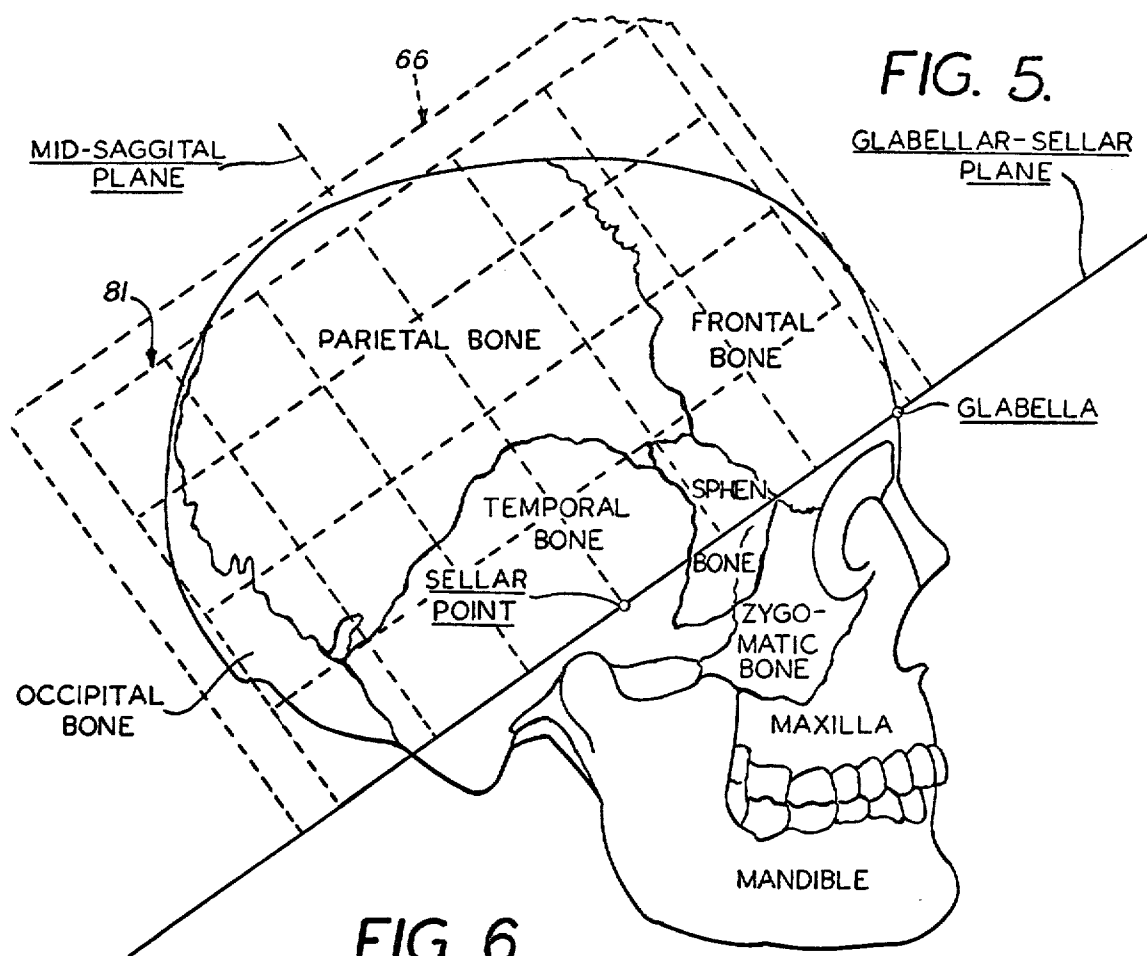
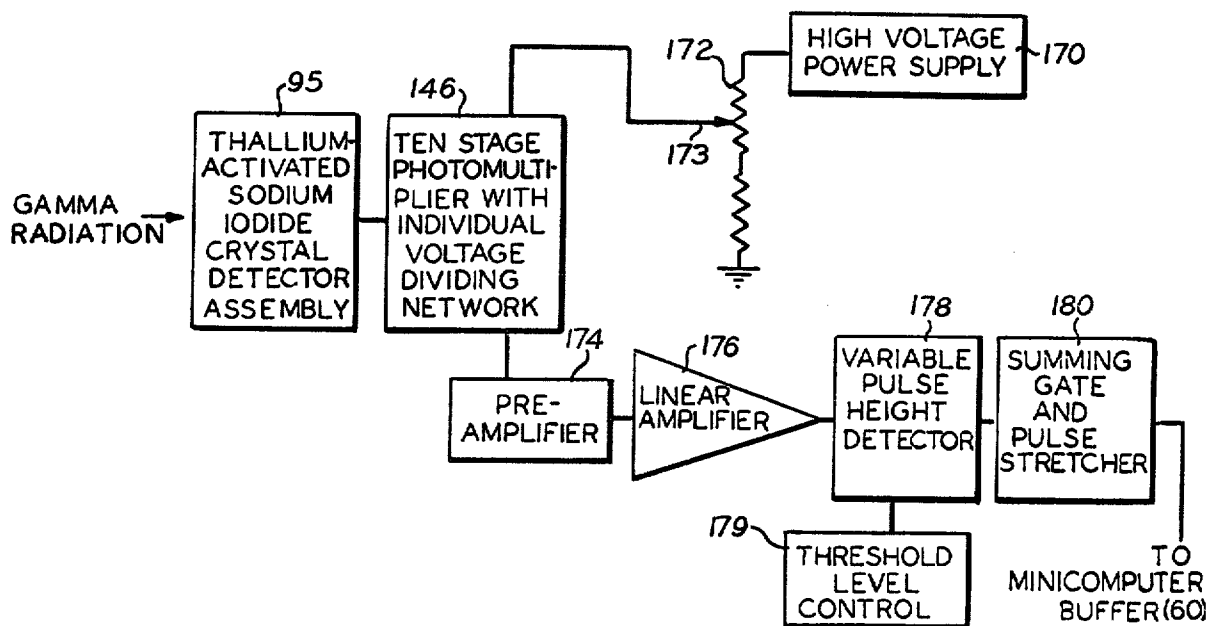


FIG. 6.



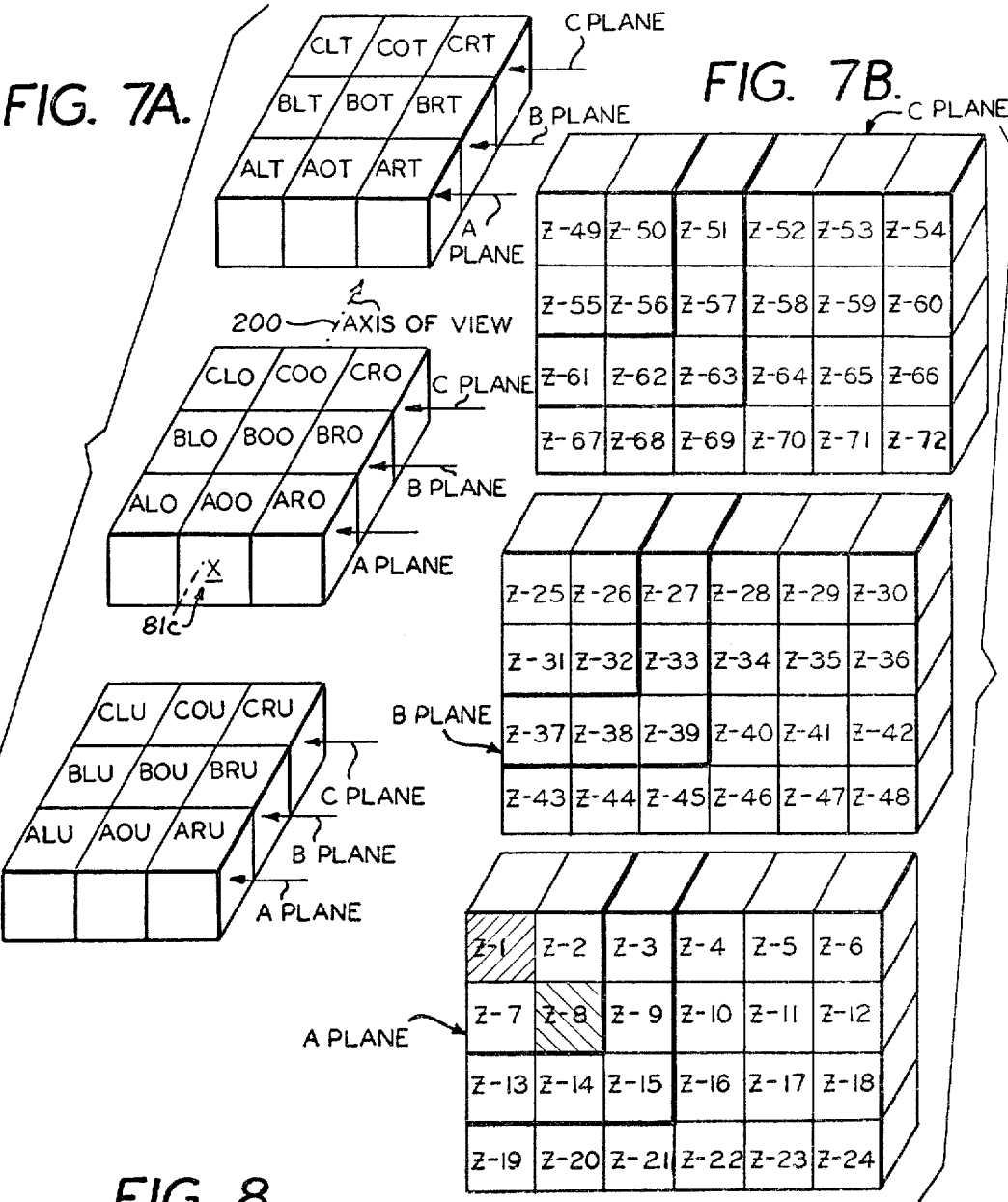


FIG. 8.

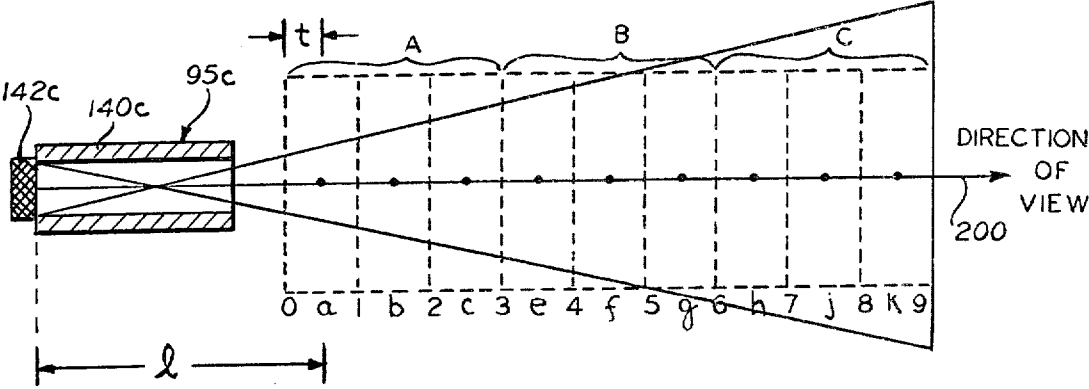


FIG. 9A.

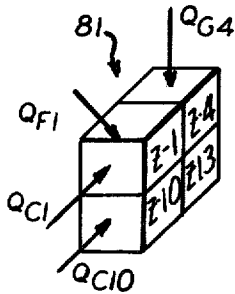


FIG. 9B.

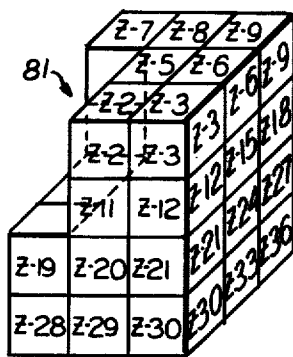


FIG. 9C.

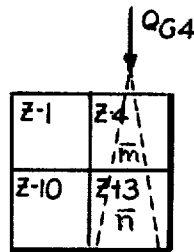


FIG. 9D.

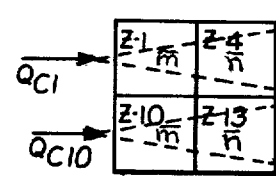


FIG. 9E.

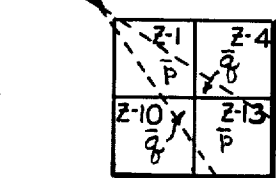


FIG. 10C.

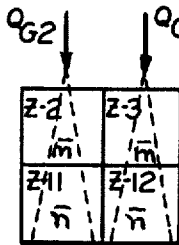


FIG. 10A.

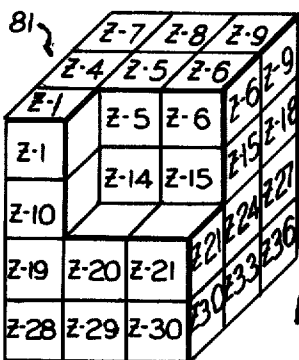


FIG. 10B.

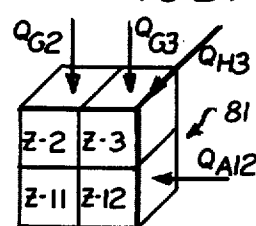


FIG. 10D.

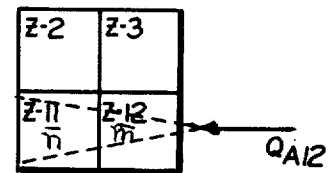


FIG. 10E.

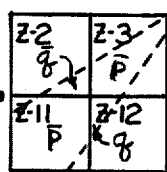


FIG. 11C.

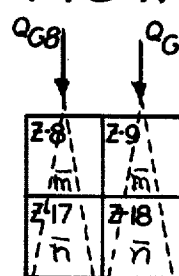


FIG. 11D.

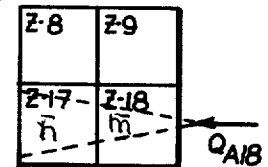


FIG. 11A.

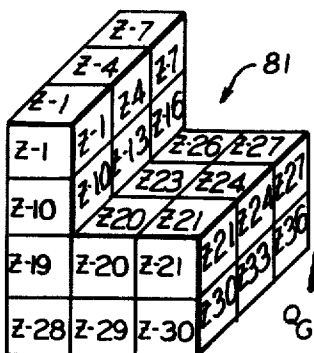


FIG. 11B.

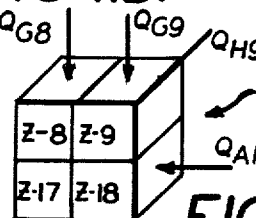


FIG. 11E.

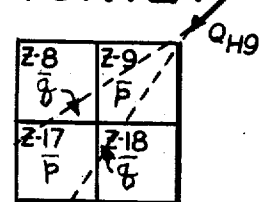


FIG. 12B.

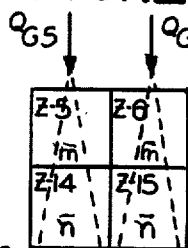


FIG. 12C.

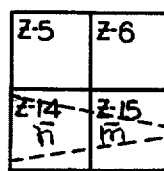


FIG. 12A.

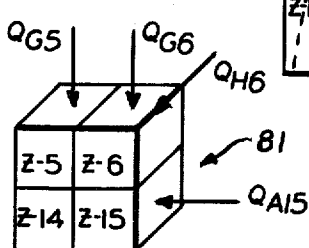


FIG. 12D.

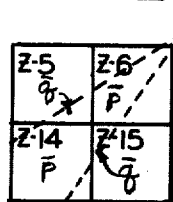


FIG. 13A.

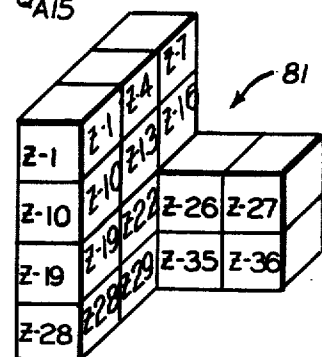


FIG. 13B. FIG. 13C. FIG. 13D. FIG. 13E.

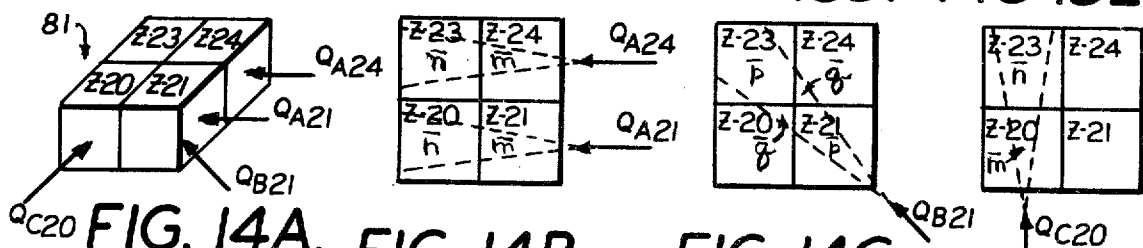


FIG. 14A. FIG. 14B. FIG. 14C. FIG. 14D.

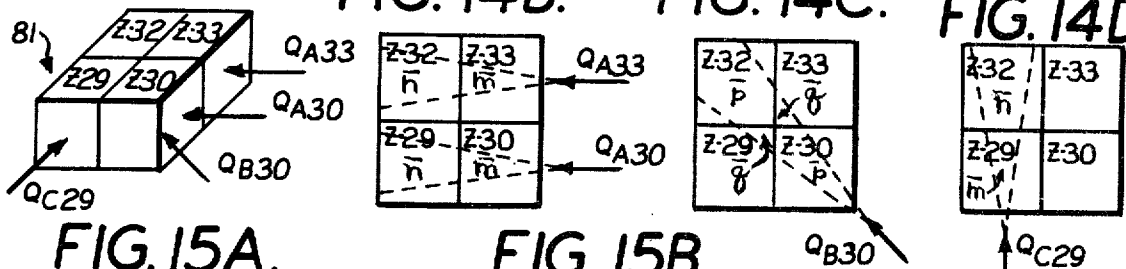


FIG. 15A.

FIG. 15B.

FIG. 16A.

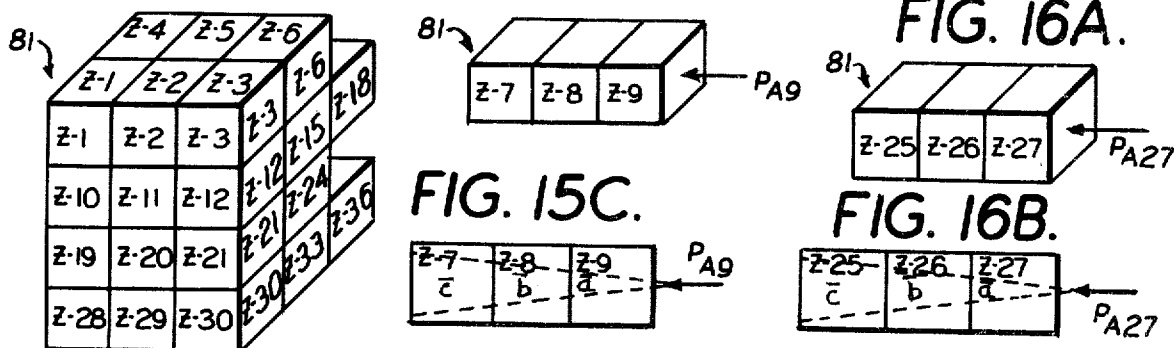


FIG. 17A.

FIG. 17B.

FIG. 18A.

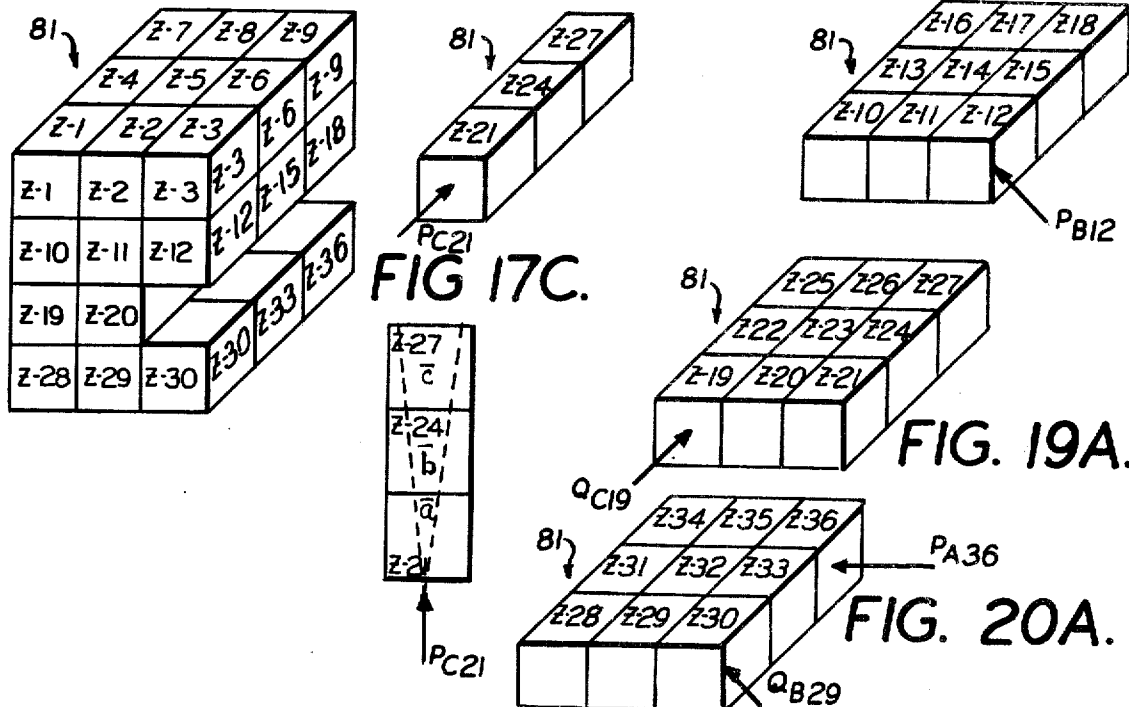


FIG. 18B.

Z-16 h	Z-17 xg	Z-18
Z-13 g	Z-14 f	Z-15 e
Z-10	Z-11 e	Z-12 d

FIG. 19B.

Z-25	Z-26	Z-27
Z-22 h	Z-23	Z-24
Z-19 m	Z-20	Z-21

FIG. 20B.

Z-34 c	Z-35 b	Z-36 a
Z-31	Z-32	Z-33
Z-28	Z-29	Z-30

FIG. 20C.

Z-34	Z-35	Z-36
Z-31 p	Z-32 q	Z-33
Z-28 o	Z-29 p	Z-30 k

FIG. 21B.

Z-11 a	Z-4	Z-7 a
Z-10 b	Z-13	Z-16 b
Z-19 c	Z-22	Z-25 e
Z-28 m	Z-31 n	Z-34

FIG. 22B.

Z-2 k	Z-5	Z-8
Z-11 d	Z-14 e	Z-17
Z-20 e	Z-23 f	Z-26 g
Z-29	Z-32 g	Z-35 h

FIG. 21A.

Z-1	Z-4	Z-7
Z-10	Z-13	Z-16
Z-19	Z-22	Z-25
Z-28	Z-31	Z-34

FIG. 22A.

Z-2	Z-5	Z-8
Z-11	Z-14	Z-17
Z-20	Z-23	Z-26
Z-29	Z-32	Z-35

FIG. 23A.

Z-3	Z-6	Z-9
Z-12	Z-15	Z-18
Z-21	Z-24	Z-27
Z-30	Z-33	Z-36

Z-3 k	Z-6	Z-9
Z-12 d	Z-15 e	Z-18
Z-21 e	Z-24 f	Z-27 g
Z-30	Z-33 g	Z-36 h

FIG. 23B.

METHOD FOR DETERMINING DISCRETE LOCALIZED BLOOD FLOW IN A LARGE VOLUME OF TISSUE

CROSS-REFERENCE TO RELATED APPLICATIONS

This application is a continuation-in-part of U.S. Ser. No. 126,770, filed Mar. 22, 1971, now U.S. Pat. No. 3,769,966, and entitled "Apparatus for Determining Local Cerebral Blood Flow".

BACKGROUND OF INVENTION

1. Field of the Invention

The present invention relates to a method for determining localized blood flow in large volumes of tissue such as local cerebral blood flow.

2. Description of the Prior Art

The ability to measure localized blood flow in various organs of the body such as cerebral blood flow within the human brain is an important medical diagnostic tool as variations in a normal flow, such as in the brain, can indicate actual, probable, or impending cerebral medical deficiencies, such as possible stroke or brain trauma. Since the human brain is the vital command center for the body, it is important that any possible malfunctions which affect the brain be discovered as rapidly as possible. Prior art diagnostic techniques for determining total or regional cerebral blood flow have involved the use of a radioactive diagnostic agent dissolved in the blood. However, such prior art techniques have only been able to determine a total or regional blood flow, which is a mean rate of blood flow through a region, such region normally being defined by a large "cone" of tissue. One such prior art device which might provide a means of estimating a qualitative indication of regional blood flow utilizes a uniplanar detector array. However, such a device has a poor energy and spatial resolution and cannot detect small blockages of blood flow within the brain; even large reductions in local flow are not easily discernible.

Other prior art techniques of this type utilize a multiplicity of probes in a random array to "view" a large randomly located cone shaped region in the brain for determining a mean rate of cerebral blood flow in this single random location. However, no provision is made for relating the amount of radioisotope contained within the intercepted tissue with the observed count rate in a quantitative manner. In addition, if there is a variation in cerebral blood flow from that of a previous measurement, there is no way of pinpointing the exact location of the flow reduction other than that there is one in the overall region.

Some of these prior art techniques have utilized the inhalation or intra arterial injection of a radioisotopic gas as the diagnostic agent in conjunction with a regional blood flow measurement. However, the prior art inhalation technique requires the patient to breathe the radioactive gas until equilibration and subsequent elimination occurs. The equilibration period is normally seven to ten minutes using a closed breathing circuit of a device such as a basal metabolism or anesthesia machine followed by a "washout" period of greater than 20 minutes. The washout of the radioisotopic gas is then measured with an external radiation detector directed toward the region. However, once again, only a mean rate of blood flow, or regional blood flow,

through the region, or cone of tissue if that is the defined region, is provided.

The technique wherein the radioactive diagnostic agent is injected into the internal carotid artery is considerably more dangerous to the patient and cannot be used except under serious medical indications for the test. While this latter technique obviates the contribution of extracerebral radioisotope, the time limitations for measurement are equally long and the method still only provides a regional blood flow measurement.

One serious limitation of these prior art regional blood flow determination methods, each of which utilize radioisotopic indicators, is that the extra cranial radiation detectors measure radiation activity from an imprecisely defined cone of tissue without being able to distinguish the depth from which the radiation activity emanates. Therefore, quantization of blood flow in brain substructures within the cone is not possible other than in the cerebral cortex which can be monitored with tangentially placed detectors. However, since nonuniformity in the distribution of cerebral blood flow during health as well as during disease is normally present, there being as much as ten to twenty-fold difference between blood flow in the cortical white matter and in the inferior colliculus, or, during periods of hyperperfusion or hypoperfusion, the various parts of the brain manifesting strikingly different changes in the associated blood flow, some changes being large while others being quite small, such prior art regional techniques have not been satisfactory as a diagnostic tool, such as in attempting localization of local cerebral ischemia, the essential lesion of stroke. Furthermore, these prior art techniques have not been satisfactory where quantitative assessment of blood flow and ultimately of metabolic rate in many specific loci of the brain simultaneously is desired.

Since these prior art techniques are slow, of low resolution, and provide only generalized regional mean rate blood flow data as opposed to localized, or direct discrete measurements of localized blood flow in a large volume of tissue, such as cerebral blood flow in a defined discrete localized portion of the brain, these techniques have not gained wide acceptance for other than cases in which such generalized data is acceptable.

These disadvantages of the prior art are overcome by the present invention.

SUMMARY OF THE INVENTION

A method is provided for determining discrete localized blood flow in a large volume of tissue, such as local cerebral blood flow in a discrete defined radiation emission detection zone of a patient's brain. The brain is subdivided into a plurality of detection zones, which are preferably identical, by means of a headgear apparatus which defines these detection zones. The headgear apparatus is preferably aligned on the patient's head with respect to three mutually perpendicular reference planes; namely, the glabellar-sellar plane as a reproducible basal plane, the mid-sagittal plane as a zero reference plane, and a reference plane tangent at the forehead. This headgear apparatus preferably includes an array of 144 or more radiation emission detectors arranged in a multiplanar array about the brain so as to provide an equal number of defined viewing or detection zones in the brain. The detectors, such as semiconductors, thalliumactivated sodium iodide crystals, ultrapurified germanium detectors or proportional

counter tubes each have associated therewith a collimator which has a lumen of predetermined configuration, preferably square, in order to present a predetermined portion of the plurality of detection zones to the associated detector.

A radioisotopic gas, such as one having two gamma energy levels, such as Xe^{133} or Kr^{85} , or a gaseous mixture of two different gamma energy level isotopes, is inhaled by the patient in a closed inhalation-exhalation breathing system over a predetermined data collection period, such as 1 minute. During this period, the concentration of this gas in the blood is increased due to periodic injection of a specified additional amount of the gas into the breathing system. Measurements are made of the arterial blood radioisotope concentration, such as by a means separate from the 144 detectors, which determines the radioisotope concentration of the gas in the last part of an exhalation of the patient during the data collection period. A determination is also preferably made of the quantity of radioisotope concentration associated with each discrete detection zone based upon measurements of resultant radiation emission by the detector array, such as by detecting both gamma energy levels, the field of view of the detector depending on the gamma energy level being detected, and a predetermined relationship between the plurality of detection zones which provides a plurality of solvable simultaneous equations relating the contributions of various detection zones in the plurality to each quantified resultant radiation emission. These equations are solved to provide the tissue radioisotope concentration for each discrete detection zone. Of course, if desired, only one gamma energy level can be detected and the radioisotope concentration can still be determined; however, the detection of two gamma energy levels having different depths of penetration facilitates the obtaining of the desired results and is preferred over the use of single energy level detection, at least with respect to the detection of cerebral blood flow. The local cerebral blood flow for a discrete detection zone is provided from the tissue and arterial radioisotope concentrations for the zone, which concentrations are preferably repeatedly determined at 10 second intervals during the 1 minute data collection period, in accordance with another predetermined relationship between these factors and cerebral blood flow, such as by applying the well known Fick-Kety expression.

BRIEF DESCRIPTION OF DRAWINGS

FIG. 1 is a diagrammatic illustration of the preferred system for use in the method of the present invention;

FIGS. 2A and 2B are diagrammatic illustrations of the headgear apparatus portion of the system for use in the method of the present invention illustrating an alternative arrangement of detectors for detecting single gamma energy radiation emission in accordance with the method of the present invention;

FIG. 3 is a block diagram of the radiation detection signal processing circuitry portion for one detection channel of the headgear apparatus;

FIG. 4 is a graphical illustration of isotopic concentration utilized in describing the method of the present invention;

FIG. 5 is a diagrammatic illustration of the human skull from the side utilized in explaining the preferred orientation of the headgear apparatus for use in the method of the present invention;

FIG. 6 is a block diagram partially in schematic, similar to FIG. 3, of an alternative embodiment of the signal processing circuitry for one detection channel;

FIG. 7A is an exploded perspective view of a mathematical model utilized in describing an alternative method of the present invention wherein a single gamma energy level is detected;

FIG. 7B is an exploded perspective view of a mathematical model, similar to FIG. 7A, utilized in describing the alternative method of the present invention;

FIG. 8 is a diagrammatic illustration utilized in explaining the field of view of a single detector in accordance with the alternative method of the present invention;

FIGS. 9A-9E is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of one gamma energy level in a two-by-two portion of a typical quadrant of the detection zone array;

FIGS. 10A-10E is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 9A-9E in a different two-by-two portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 11A-11E is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy as in FIGS. 9A-9E in still a different two-by-two portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 12A-12D is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 9A-9E in another two-by-two portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 13A-13E is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 9A-9E in still another two-by-two portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 14A-14D is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 9A-9E in a different three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 15A-15C is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of a different gamma energy level than in FIGS. 9A-9E in a three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 16A and 16B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in a different three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 17A-17C is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in still a different three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 18A and 18B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in a different three-by-three portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 19A and 19B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 9A-9E in a different two-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 20A-20C is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of both the gamma energy levels of FIGS. 9A-9E and FIGS. 15A-15C in another three-by-three portion of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 21A and 21B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in two different three-zone-deep portions of the same typical quadrant as in FIGS. 9A-9E;

FIGS. 22A and 22B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in a different substantially three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E; and

FIGS. 23A and 23B is a mathematical model utilized in describing the preferred method of the present invention illustrating the detection of the same gamma energy level as in FIGS. 15A-15C in still a different substantially three-zone-deep portion of the same typical quadrant as in FIGS. 9A-9E.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

General Description

Referring now to the drawings in detail and especially to FIG. 1 thereof, the system of the present invention, generally referred to by reference numeral 10, for practicing the method of the present invention is shown. The preferred method of the present invention shall be described, by way of example, for determining local cerebral blood flow, which is the blood flow within a discrete preselected portion of a patient's brain although the preferred method of the present invention may be utilized, if desired, to determine discrete localized blood flow in any large volume of tissue such as in the myocardium, the lung, the kidney, the liver, the spleen or any other body organ, or to resolve the metabolic rates of component regions of the brain and other organs.

It should be understood that FIG. 1 is merely a diagrammatic illustration of the system 10 for purposes of explanation, and is not meant to indicate the critical dimensions of the system, such as preferred requirement of minimum dead space in the closed breathing system or the critical arrangement and orientation of radiation detectors in the headgear apparatus. The system 10 of the present invention preferably includes a breathing mask 12 which is preferably removably mounted on the face of the patient during the performance of the local cerebral blood flow determination so as to isolate the breathing passages of the patient from the surrounding

environment and provide a closed inhalation-exhalation breathing system, which will be described in greater detail hereinafter. The mask 12, which is preferably a conventional breathing mask, is preferably formed of a material which is pliable enough to provide a comfortable but close fit on the face of the patient in the mounted position so as to reduce apprehension during a data collection period which is associated with the local cerebral blood flow determination.

The mask 12 includes a conventional type of mouthpiece breathing port 14 which is provided with an inlet or inhalation conduit 16 and an outlet or exhalation conduit 18. Conduits 16 and 18, respectively, supply the substance or gas to be inhaled to the mask 12 and remove the substance exhaled from the mask 12. Preferably, the inhalation conduit 16 and the exhalation conduit 18 each include a conventional demand valve 20 and 22, respectively, disposed within the conduit flow path so as to close off the respective associated conduits 16 or 18 unless the patient is either demanding the next breath (inhalation) or exhaling his previous breath (exhalation).

Inhalation demand valve 20 is preferably interposed in inhalation conduit 16 between the mouthpiece breathing port 14 and a small gas reservoir radiation-shielded spirometer or inhalation chamber 28 from which the substance or gas to be breathed by the patient is provided. Exhalation demand valve 22 is preferably interposed in exhalation conduit 18 between the mouthpiece breathing port 14 and an exhalation chamber 30 which is preferably equal in volume to the inhalation chamber 28. The inhalation chamber 28 and the exhalation chamber 30 are preferably connected together in the closed inhalation-exhalation breathing system through a conventional gas purification unit 32 having an inlet conduit 34 connected to the exhalation chamber 30 outlet, and an outlet conduit 36 connected to the inhalation chamber 28 inlet. The gas purification unit 32 operates in a conventional manner to remove excess carbon dioxide from the breathing system, add oxygen to the breathing system, and filter the air passing therethrough. The gas purification unit 32 is preferably operated responsive to the beginning of a breath by the patient in a manner to be described in greater detail hereinafter.

In practicing the preferred method of the present invention, as will be explained in greater detail hereinafter, in order to introduce a diagnostic agent into the cerebral blood, the patient breathes a radioactive gas which most preferably has two distinct gamma energy levels such as the radioisotopic gas Xenon¹³³, which is the preferred diagnostic agent, although a gaseous mixture of two different radioisotopic gases, each having different gamma energy levels could be utilized as the diagnostic agent. In addition, if desired, alternatively, a radioisotopic gas having only a single gamma energy level or one having more than one level with only one level being detected could be utilized if the improved resolution of the above is not desired. As will also be explained in greater detail hereinafter, a high concentration of this gas is desirable during the performance of the local cerebral blood flow determination and, most preferably, this concentration increases during the quantification of the data necessary for the local cerebral blood flow determination. For purposes of explanation, the preferred method of the present invention, as well as an alternative method thereto shall be de-

scribed with the use of radioactive Xenon¹³³ gas as the diagnostic agent.

In order to accomplish the previously described desired condition, a radioactive gas injection system 40 is connected through the valve 42 via a conduit 44 to the inlet of the exhalation chamber 30 for injecting a predetermined quantity of the radioactive Xenon¹³³ gas into the exhalation chamber 30 each time valve 42 is opened. In addition, gas injector 40 also preferably includes a conduit 48 which is connected to the inlet of the inhalation chamber 28 via another valve 50 which is preferably only opened at the end of the patient's next breath after the beginning of the data collection period so as to also inject this predetermined quantity of radioactive gas into inhalation chamber 28 as well as exhalation chamber 30 at this time. Preferably, valve 50 remains closed throughout the remainder of the data collection period and gas injector 40 only injects the radioactive gas into the exhalation chamber 30 where it is mixed, such as by an impeller (not shown), with the gas contained therein and passed through the gas purification unit 32 to the inlet of the inhalation chamber 28. If desired, other variations of this system may be utilized, such as by locating the purifier 32 between valve 22 and chamber 30, so that fresh radioisotope need not pass through purifier 32.

Preferably, for purposes of gas evacuation at the completion of the data collection period, another valve 52 is interposed in inhalation conduit 16 between demand valve 20 and the mouthpiece breathing port 14, and another valve 54 is interposed in conduit 18 between the exhalation demand valve 22 and the mouthpiece breathing port 14. Valves 52 and 54 are each, respectively, connected to conduits 56 and 58. Conduit 56 is preferably open to the surrounding environment to provide room air therethrough, and conduit 58 preferably opens to an exhaust hood (not shown) for evacuating the gas from the system 10. Valve 52 preferably blocks conduit 56 during the data collection period while leaving conduit 16 open to demand valve 20 and, similarly, valve 54 blocks conduit 58 during this period while leaving conduit 18 open to demand valve 22. As will be explained in greater detail hereinafter, at the conclusion of this period, valves 52 and 54 are operated, such as electrically via a signal from a conventional programmable minicomputer 60, such as a Honeywell 316, which may be utilized to coordinate the data collection and the resultant cerebral blood flow determination. The actuation of valves 52 and 54 changes them from their normal mode to the gas evacuation mode wherein conduits 16 and 18 are closed off to demand valves 20 and 22 and inhalation chamber 28 and exhalation chamber 30, respectively, and open to conduits 56 and 58, respectively.

In this manner, a closed inhalation-exhalation breathing system is provided during the data collection period for the local cerebral blood flow determination from inhalation chamber 28, via inhalation conduit 16 to mouthpiece breathing port 14 of the mask 12, via mouthpiece breathing port 14 to exhalation conduit 18, via exhalation conduit 18 to the exhalation chamber 30, via exhalation chamber 30 to the gas purification unit 32, and via gas purification unit 32 back to the inhalation chamber 28; with the radioactive gas being injected periodically during this period into the closed inhalation-exhalation breathing system from gas injector 40.

The portion of the system utilized in making a quantitative local cerebral blood flow determination will now be described. As will be explained in greater detail hereinafter, a conventional general purpose computer 61, such as an IBM 360, may preferably be utilized in conjunction with, or in place of, the minicomputer 60 in order to process the data collected during the data collection period, if desired, in order to enhance the speed in which results are obtained. However, if desired, any other conventional method of processing such data may be utilized. For purposes of illustration, it shall be assumed that the general purpose computer 61 is utilized in conjunction with minicomputer 60 to process this data.

A conventional end-tidal sampler 62, such as a Falk-Kupferman end-tidal sampler, preferably has an inlet which is connected to exhalation conduit 18 for sampling the alveolar or end-tidal gas concentrations at the end of each breath, and an outlet which is connected to a radioactive gas analyzer 63, to be described in greater detail hereinafter, which preferably includes a coiled passageway 64 of known volume and a radiation emission detector assembly 65. As will be described in greater detail hereinafter, the end-tidal sampler 62 contains a relay mechanism (not shown) which is normally actuated at the beginning and end of each breath in a conventional manner. This relay is preferably utilized to activate the gas injector 40 at the end of each breath and the gas purification unit 32 at the beginning of each breath in a manner to be described in greater detail hereinafter.

The end-tidal samples passed through end-tidal sampler 62 are preferably drawn through the coiled passageway 64, or another type of sampling chamber if desired; of analyzer 63, which is preferably shielded against external radiation. For efficiency and convenience, the coiled passageway 64 and detector assembly 65 are preferably located near the patient's head adjacent a headgear apparatus 66 which is mounted in a predetermined position on the patient's head preferably by means of a universal mounting means 68. A more detailed description of a typical headgear apparatus 66 will be presented hereinafter in connection with FIGS. 2A and 2B, although the detectors illustrated therein are illustratively arranged and oriented with respect to their associated fields of vision in accordance with the alternative method of the present invention, the arrangement, orientation and field of view of these detectors differing in accordance with the preferred method of the present invention, as will be described in greater detail with reference to FIGS. 9-23.

Suffice it to say at this point that the headgear apparatus 66 preferably provides radiation emission detection signals which are each fed through an associated radiation detection signal processing circuit 67, one circuit being provided for each detector although only one such circuit and signal path are shown for purposes of clarity. Circuit 67 will be described in greater detail hereinafter. The output of circuit 67 is in turn preferably fed to the minicomputer 60, which, preferably acts as a buffer in order to more efficiently handle the data collected during the data collection period as considerable expense is involved in utilizing a large scale general purpose computer 60 such as the IBM 360 "on-line" throughout the data collection period. However, if this is not a consideration, the minicomputer buffer

60 may be omitted and the data fed directly to the general purpose computer 61.

The radiation emission detector assembly 65, which is to be described in greater detail with reference to the headgear apparatus 66 which preferably contains similar assemblies, is mounted in analyzer 63 in relation to coiled passageway 64 so as to quantify the radiation emission associated with the sampled gas contained in the coiled passageway 64. Detector assembly 65, which preferably provides a signal output in response to a gamma radiation input, is connected to a radiation detection signal processing circuit 74, which is preferably similar to detection circuit 67, whose output is in turn also preferably connected to the minicomputer buffer 60.

Preferably, the gas analyzer coiled passageway 64 is also operatively connected, in a manner to be described in greater detail hereinafter, to an end-tidal sample analyzer system 75, to be described in greater detail hereinafter, in order to determine the concentrations of carbon dioxide and oxygen in the patient's breath. Both of these substances, and especially carbon dioxide, are influential upon the rate at which the blood flows through the human brain and should, therefore, preferably be taken into consideration. Normally, the patient's respiration is adjusted to maintain these gases at fairly constant levels. However, some patients may have additional diseases which may influence the level of these gases; while other patients may over-breathe during the data collection period because of tension or apprehension induced by the data collection procedure. If the patient over-breathes during this period and thus reduces his carbon dioxide pressure, his cerebral blood flow will be slowed and of an altered pattern even though his cerebrovascular system is normal. However, the relationship between the reduction in this blood flow for a given reduction in carbon dioxide pressure is known and can be easily compensated for in the final determination of cerebral blood flow if this reduction is known. Similarly, a patient having impaired ventilation may have a higher rate of cerebral blood flow which can also be predicted from the elevated carbon dioxide pressure if it is known. In addition, low arterial oxygen tension during impaired ventilation/profusion can also cause a faster rate of cerebral blood flow. It is therefore both preferable and desirable to know the carbon dioxide and oxygen pressures in order to compensate for their effects on the cerebral blood flow determination or errors may be introduced in the final determination.

The end-tidal sample analyzer system 75, preferably includes a conventional through-flow carbon dioxide gas analyzer 76, such as a Capnograph which is an infrared carbon dioxide gas analyzer, and a conventional through-flow oxygen analyzer 77, such as a Rappox manufactured by Instrumentation Associates, which analyzers 76 and 77 are preferably connected together in a serial through-flow path, although they may be connected in parallel through-flow paths if desired. The inlet to the carbon dioxide gas analyzer 76 is preferably connected to the coiled passageway 64 outlet via a conduit 78 having a valve 79 disposed therein for preferably directing the end-tidal sampled gas leaving the coiled passageway 64 either through conduit 78, from which it is drawn through the carbon dioxide analyzer 76 and therefrom through the oxygen analyzer 77 from where it is exhausted to the closed circuit breathing sys-

tem or through another conduit 80, through which it is exhausted directly to the exhaust hood. Preferably, the valve 79 only passes the end-tidal sampled gas through the analyzer system 75 prior to the initiation of the actual data collection period associated with the local cerebral blood flow determination, the valve 79 directing the gas through conduit 80 during this period, in order to quantify or measure the oxygen and carbon dioxide gas pressures present in the end-tidal sample as well as the period of time in which the carbon dioxide gas present in the end-tidal sample is on the plateau portion of its exhalation curve, which curve is similar to that for Xenon¹³³ gas. The end-tidal sampler 62 may then preferably be normalized or calibrated to operate on this portion of the exhalation curve during the data collection period.

Furthermore, the oxygen and carbon dioxide pressure information may be utilized to determine the time at which the actual data collection period is to commence, which is preferably when these pressures are stabilized. At that time, the data collection may proceed without introduction of any significant errors in the cerebral blood flow determination due to carbon dioxide or oxygen gas pressure variations. If desired, these measurements may be displayed visually and the data collection begun manually by an operator by actuation of a start switch (not shown) when these pressures are stabilized, or in the alternative, these values may generate corresponding electrical signals which when fed to a comparator (not shown) may provide a "begin data collection" signal to the minicomputer 60 at stabilization if the minicomputer 60 is utilized as an overall control mechanism for the data collection procedure.

As will be explained in greater detail hereinafter, the normalized end-tidal sampler 62 is preferably initially activated from minicomputer 60 at the time when the actual data collection period is to commence so as to synchronize the selection and radiation emission detection times of the end-tidal gas samples with the radiation emission detection times of the headgear apparatus 66, the end-tidal sampler 62 being activated in a conventional manner thereafter. If the start signal is manually initiated, then this signal merely puts the minicomputer 60 "on-line" so that it may provide the end-tidal sampler 62 synchronization signal; whereas, if the minicomputer 60 is utilized as an overall control, then it can be programmed to only provide this synchronization signal after it has received the start signal from the comparator.

If desired, the carbon dioxide and oxygen gas analyzers 76 and 77, respectively, may be electrically connected to the minicomputer 60 in order to provide dynamic pressure information during the data collection period instead of, for example, merely providing a start signal and, in this instance, the computer 61 may then be preferably programmed to compensate for blood gas tension variations in the resultant local cerebral blood flow determination. In such instance, the valve 79 would be open so as to direct the end-tidal gas sample to the calibration system 75 throughout the actual data collection period, as opposed to only prior thereto.

HEADGEAR APPARATUS

Now referring to FIGS. 2A and 2B and describing the headgear apparatus 66 in greater detail, the system 10 of the present invention, as will be explained in greater

detail hereinafter, preferably "subdivides" the patient's brain into a predetermined plurality of discrete detection zones for purposes of determining the rate of cerebral blood flow in each of these zones, although, if desired, the final determination may be limited to any number of zones in the plurality, such as one. These detection zones, which are shown by way of example by dotted lines in FIGS. 2A and 2B are preferably represented by three-dimensional cubes **81-81** which are each of equal volume and dimensions. It should be noted that these detection zones **81** are merely mathematical models of sub-volumes defined within the overall brain volume by means of the headgear apparatus **66** and are preferably correlated to a "brain map" so as to pinpoint their exact location within the brain. Most preferably, 144 such detection zones are utilized to define the overall brain volume, although other numbers might be employed. As will be explained in greater detail hereinafter, during the data collection period, each of these zones **81** will be provided with an associated radioisotope concentration and corresponding radiation emission due to the presence of the radioactive gas in the cerebral blood. Hence, these zones **81** are termed radiation emission detection zones.

As was previously mentioned, the arrangement and orientation of the detectors, which will be discussed in greater detail hereinafter, illustrated in FIGS. 2A and 2B, and their associated illustrated fields of view, are in accordance with an alternative method of the present invention, to be described in greater detail hereinafter, wherein only a single gamma energy level of the radioisotopic gas is detected. Before describing this alternative method and arrangement which differs from the preferred arrangement utilized in accordance with the preferred method of the present invention, the preferred arrangement shall be described with reference to FIGS. 9-23 wherein two different gamma energy levels of the radioisotopic gas are detected, these levels, by way of example, being 31 Kev and 81 Kev for Xenon ¹³³.

As will be explained in greater detail hereinafter, the headgear apparatus **66** should preferably bear a specific and reproducible relationship to the patient's brain. Therefore, in aligning the headgear apparatus **66**, a reproducible basal plane, which defines the angle of viewing to which all other planes will be parallel, perpendicular, or angulated at 45 degrees, must be chosen. In the preferred system **10** of the present invention the glabellar-sellar plane (see FIG. 5), which is the plane which delimits the lower surface of the human forebrain, is preferably chosen as a reference plane. The mid-sagittal plane (see FIG. 5), which is the plane perpendicular to the basal plane and midway between its lateral limits, is preferably chosen as the second plane of orientation or reference in the preferred alignment position of the headgear apparatus **66**. The third plane of orientation or reference for the headgear apparatus **66**, which thereby defines the exact alignment or orientation of the headgear apparatus **66** with respect to the brain of the patient, is the plane perpendicular to the basal plane and the mid-sagittal plane and tangent to the forehead (see FIG. 5). When the headgear apparatus **66** is aligned in these three planes, with the mid-sagittal plane preferably being the zero reference plane, the patient's brain volume is subdivided into the plurality of detection zones **81**, which are preferably cubic or rectangular parallelepiped blocks in volume, arranged in a rectangular six-by-six array in

the glabellarsellar plane and four blocks high in the mid-sagittal plane, as shown in FIG. 5 by the dotted lines.

Referring now to FIGS. 9-23, the array of detectors is preferably arranged to view a block-shaped volume (FIG. 5) of sufficient size to include a typical adult human head although, if desired, the size could be varied to include any size head such as a child's. The headgear apparatus **66** preferably includes a substantially rectangular hollow helmet housing **82** having a front portion **84**, a rear portion **86**, a top portion **88**, side portions **90** and **92**, and a bottom portion **94** through which the head is inserted, the housing **82** forming a rectangular parallelepiped. By way of example, and not limitation, typical internal dimensions for a typical housing **82** which should accommodate approximately 100 percent of the adult female heads and approximately 99 percent of adult male heads are approximately 8½ inches from anterior to posterior, 6¾ inches from side to side, and 5 21/32 inches from superior to inferior, the total volume being subdivided into 144 smaller rectangular blocks or detection zones **81** of equal volume, each being, in the example given above, preferably 36×28.6×36mm. The large rectangular block encompassing the brain volume is preferably subdivided into six divisions from side to side, six from front to back and four from top to bottom, resulting in the 144 detection zones **81** preferably requiring, as will be described in greater detail hereinafter, the solution of 144 unknowns representing the quantity of radioisotope in each of the 144 zones **81**, an equal number of detectors (144) preferably being provided spaced about the brain volume within the helmet housing **82**.

Preferably, the radioisotopic gas, such as Xenon ¹³³, has a limited penetration for the gamma photons emitted thereby which enables the 144 block or zone **81** volume to be preferably divided into four equal quadrants of 36 blocks each, each quadrant being 3×3×4 blocks, for purposes of determining the local cerebral blood flow in each detection zone **81** in accordance with the preferred method of the present invention, **36** detectors being associated with each quadrant. Of course, if desired, the entire 144 zone block may be treated as a whole rather than as four equal quadrants, although when the quadrants can be treated separately (depending on gamma energy) as preferred errors in measurement which may inadvertently occur in one quadrant cannot affect the results in the other quadrant. Preferably, the block or zone sizes are chosen so that natural attenuation will limit the source of the lower gamma energy (31 Kev for Xenon ¹³³) photons almost entirely to the two superficial zones in depth of the array and the higher gamma energy (81 Kev for Xenon ¹³³) photons to three zones in depth (skull and scalp attenuation included). The presently preferred arrangement of detectors on a typical 36 zone quadrant (zones Z-1 through Z-36) is illustrated in FIGS. 9-23 wherein arrows labeled with Q and P subscripts, respectively, represent the position of the low energy and high energy detectors, respectively, in the detector array, the illustrated quadrant representing the front-right quadrant.

As will be described in greater detail, each detector is preferably fabricated from p type high purity germanium and has an associated collimator to define the field of view illustrated in FIGS. 9-23 by the dotted lines for the Q or low energy level detectors and the P

or higher energy level detectors. Preferably each of the detectors has an associated single channel signal analyzer (see FIG. 3), each detector channel preferably having the capability of quantitating and discriminating more than one photon or gamma energy level within its field of view, the energy of the photon determining its tissue penetration. For purposes of explanation, separate detectors are illustrated (Q and P) for low energy and high energy photons, the associated field of view being selected in accordance with the setting of the single channel analyzer window; however, if desired, detectors may be utilized which quantitate both photon energies simultaneously. Preferably, a plurality of individual detector assemblies (similar in structure to the alternative detector illustrated in FIG. 8 but differing in field of view) equal in number to the plurality of radiation emission detection zones 81, which in the preferred method of the present invention is 36 for each quadrant, are arrayed about each quadrant of the housing in various orientations as illustrated by the P and Q arrows in FIGS. 9-23. Each of the detector assemblies is preferably substantially identical in structure only differing in the field of view provided by the associated collimator (approximately two zones 81 deep for the Q detectors and approximately three zones 81 deep for the P detectors as illustrated in FIG. 21B by way of example) and the setting of the single channel analyzer window to detect low energy photons (e.g. 31 Kev) for the Q group of detectors and high energy photons (e.g. 81 Kev) for the P group of detectors, the field of view for all P group detectors preferably being substantially equal in volume with the variations therefrom illustrated in FIGS. 18B, 22B and 23B, and the field of view for all Q group detectors preferably being substantially equal in volume with the variations therefrom illustrated in FIGS. 9E, 10E, 11E, 12D, 13D, 14C and 20C, each of these 36 individual detector assemblies preferably yielding an independent linear simultaneous equation.

As shown and preferred in FIGS. 9-23, each of the detector assemblies represented by the P and Q arrows is preferably located at the center of the point of "entry" into the adjacent detection zone 81 so as to be substantially symmetrical about the longitudinal viewing axis of the detector field of view. Thus, as shown and preferred in FIGS. 9-23, the P and Q groups of detector are arranged at 26 positions at the center of the associated adjacent viewing surface defined by the corresponding detection zone 81 in the quadrant and at 10 positions substantially along a 45 degree line at the center of the corner of the associated adjacent viewing surface defined by the corresponding detection zone 81 in the quadrant for a total of 36 different positions inclusive of all 36 zones 81 in the quadrant yielding the preferred 36 independent simultaneous linear equations containing all 36 unknowns representative of the quantity of radioisotope in each of the 36 zones 81 in the quadrant, this procedure being repeated for each of the other quadrants to yield the preferred 144 independent simultaneous linear equations containing all 144 unknowns representative of the quantity of radioisotope in each of the 144 zones 81 in the entire brain volume.

Preferably, as was previously mentioned, the headgear apparatus 66 is designed for the largest typical adult human head likely to be encountered and, therefore, variations in the air space between the detector assemblies 95 and the scalp of the patient may occur.

This variable air space is due to variable cranial curvatures associated with different patients as well as variations in the size of the head. In order to compensate for these variations in air space, if desired, which variations affect the quantity of the radiation emissions detected by the detector assemblies, a gauge-type means may preferably be included in the helmet housing 82 substantially at the position of each of the detector assemblies which measures the size of the air space between the associated detector assembly and the scalp of the patient. These gauges (illustrated by way of example in the alternative arrangement of FIGS. 2A and 2B by reference numerals 130, 132, 134, 136 and 138) may be calibrated rods or, if desired, may be conventional position transducers so as to provide a contemporaneous measurement of the air space between the associated detector assembly and the scalp of the patient during the data collection period. If desired, a separate head-gauge, comprising solely position transducers or calibrated rods located at each of the detector assembly positions in an identical array with that of the helmet housing 82, may be utilized prior to the use of the headgear apparatus 66 so as to provide these air space measurements prior to the actual data collection period. In this instance, the headgear apparatus 66 is preferably not provided with a gauge at each detector assembly position, but rather is preferably merely provided with only one calibrated rod or position transducer gauge in an identical position with that of such gauge in the head-gauge in each of the portions or planes of the helmet housing 84 through 92, inclusive, illustratively shown by gauges 130 through 138, inclusive, so as to insure that the helmet housing 82 is aligned in the identical position in which the head-gauge was aligned when the air space measurements were made. The universal mounting means 68, as was previously mentioned, is preferably capable of aligning the headgear apparatus 66 in this desired alignment or orientation position and preferably includes locking means (not shown) for retaining it in this position during the data collection period. If a separate head gauge device is utilized to measure the variable air space, as was previously mentioned, then this device should preferably be similarly oriented in the manner previously described with reference to the orientation of the headgear apparatus 66 prior to the taking of these measurements.

DETECTOR ASSEMBLY

Now describing the radiation emission detector assembly in greater detail, the detector assembly preferably being functionally identical with detector assembly 65 of the Xenon gas analyzer 63 as was previously mentioned except differing in the associated field of view and single channel analyzer window or setting. A typical preferred detector assembly 95a or 95c is shown in FIGS. 2A and 8, respectively, this detector assembly being functionally identical with the P and Q group of detectors except differing in the illustrated associated field of view, as was previously mentioned, and the associated single channel analyzer window or setting. Accordingly, for purposes of explanation, the structural details of the preferred detector assemblies 95, as well as the P and Q groups of detectors, all of which are preferably functionally identical, will be described hereinafter only with reference to detector assembly 95a. The detector assembly 95a preferably includes a collimator portion 140 made of radiation absorbing

material cooperatively associated with a radiation emission detector 142. Preferably, the collimator portion 140 of the detector assembly 95a is a tube of specific length and cross-sectional lumen area whose function is to limit the direction from which radiation may enter the detector 142. It should be noted that it is the collimator 140 which primarily determines the degree of spatial resolution of the system 10 of the present invention. In the preferred system 10 of the present invention, the shape of the collimator lumen 144 is preferably square so as to maximize the area of the detector 142 utilized for viewing, although other shapes, such as circular, may be utilized therefor if desired. The collimator tubing 140 is formed of a material which has a high radiation absorption characteristic, such as a composite material composed of 90 percent tungsten plus 10 percent nickel and copper (e.g. Kullite 112 or Mallory 1000), so as to sharply restrict the radiation emission to the lumen-area of the detector 142.

The length of the collimator 140 determines the volume of tissue observed at the detector 142 through the lumen 144 of a given cross-sectional area and, therefore, determines the number of detection zones 81 observed by the detector 142 in its direction of view. The preferred square collimator lumen 144 provides a square prism field of view for the detector 142, each detector 142 arrayed about the helmet housing 82 preferably looking into the face of the cubic detection zone 81 associated therewith, as previously described, and seeing its own prism of tissue. The collimator 140 length is preferably chosen so that the prismatic base of the field of view for the P group of detectors is substantially only slightly smaller in area than the area of a cube face of a detection zone 81 at the distal side of the third detection zone 81 in depth in a straight line based from the detector assembly in that direction or field of view and substantially only slightly smaller in area than the area of a cube face of a detection zone 81 at the distal side of the second detection zone 81 in depth in a straight line based from the detector assembly in that direction or field of view. In addition, with the preferred collimator 140 of the present invention, the square prism field of view for the associated detectors 142 located at an angle to the detection zone viewing surface (FIGS. 9E, 10E, 11E, 12D, 13D, 14C, 18B, 20C and 22B) preferably includes portions of the detection zones 81 in a two-by-two, four-zone matrix array for the Q group of detectors and a three-by-three, nine-zone matrix array for the P group of detectors, with the exceptions illustrated in FIGS. 22B and 23B, about the axis of view of the associated detector. The preferred collimator 140, therefore, provides a preferred effective field of view substantially three detection zones in depth for the P group of detectors and two detection zones in depth for the Q group of detectors, with a proportionate amount of such zones 81 which are in a three-by-three matrix array or two-by-two matrix array with the zone at the point of "entry" into the volume for the angled P group of detectors and Q group of detectors, respectively. Therefore, as will be explained in greater detail hereinafter, the radiation emission quantified or detected by a given detector assembly will be a resultant value dependent on the contributions to this resultant provided by the respective radiation concentrations of the radioactive gas in the portions of the detection zones 81 contained within the preferred effective field of view.

This preferred optimal choice for the collimator 140 for the P and Q groups of detectors is based upon the useful depth of view; that is, beyond a given depth of tissue, for the gamma energies of the preferred radioisotope Xe¹³³, the tissue attenuation and dispersion of radiation emission according to the inverse-square law only allows a negligible fraction of radiation emission to actually arrive at the detector 142 from these depths. Therefore, radiation emissions from beyond three detection zones 81 in depth for the P group (81 Kev) and for the Q group (31 Kev) are outside the effective fields of view as they do not significantly affect the determinations of local cerebral blood flow. If it is desired to take these negligible amounts into consideration, the collimator 140 may be designed so as to include these additional detection zones in its effective field of view.

As was previously mentioned, the detector 142 of the detector assembly 95a or the P or Q groups is preferably a semiconductor such as ultra high purity germanium. Such a solid-state detector 142 is particularly applicable to the detection of gamma radiation, such as the gamma radiation associated with the preferred gas Xenon¹³³ (31 Kev and 81 Kev) with high energy resolution, good stability and efficiency over the entire (31 Kev to 81 Kev) energy range. This is of particular concern since gamma radiation is the preferred radiation emission to be detected, beta radiation not penetrating the skull of the patient. Such a preferred solid-state detector 142 absorbs the gamma radiation and generates secondary electrons in response thereto which signal is processed in the radiation detection signal processing circuit 67 associated with each of the detector assemblies. If desired, a crystal detector, such as a crystal of thallium-activated sodium iodide may be utilized in place of the solid-state detector. However, as shown and preferred in FIG. 6, when such a crystal is utilized, a photomultiplier 146 should preferably be utilized therewith due to the fact that such a crystal emits a very weak light photon when an energy quantum of the gamma radiation is absorbed by the crystal, the photomultiplier 146 converting this light photon into electrical energy and amplifying it to a transmittable level in the form of electrical pulses. The crystal detector 142, in such an instance, is preferably interposed between the collimator 140 and the photomultiplier tube 146 so as to receive a gamma radiation emission input from the collimator 140 and provide an output to the photomultiplier tube 146 in response thereto.

SOLID STATE DETECTOR — RADIATION DETECTION SIGNAL PROCESSING CIRCUIT

Now describing in greater detail the typical radiation detection signal processing circuit 67, shown in greater detail in FIG. 3, which is preferably utilized for each of the detection channels in which the preferred solid state detector is utilized as the detector 142. As was previously mentioned, the same processing circuit 67 configuration is utilized for the Xenon¹³³ gas analyzer radiation detection signal processing circuit 74. Therefore, a detailed description of this circuit is deemed unnecessary.

Referring now to FIG. 3, the output of the solid state detector assembly, 95 for the alternative arrangement of FIGS. 2A and 2E or for the P or Q group of detectors for the preferred arrangement of FIG. 9-23, for a typical detection channel or processing circuit 67 is preferably operatively connected through a conventional

charge sensitive pre-amplifier stage 148 and conventional linear amplifier stage 150 to a conventional single channel analyzer or pulse height detector stage 152, such as a circuit comprising transistor-transistor logic (TTL), to preferably provide a TTL type output signal to the minicomputer buffer 60 which is indicative of the detected count in that detection channel, which signal is subsequently preferably fed to the general purpose computer 61. Since the three stages 148, 150 and 152 comprising the preferred radiation detection signal processing circuit 67 are preferably conventional, a detailed description is deemed unnecessary. Suffice it to say that, for the P and Q group of detectors, dependent on the energy level (31 Kev or 81 Kev) to be detected the single channel analyzer window is set for this level by conventionally gating at the selected energy level and discriminating against all other levels.

CRYSTAL DETECTOR — RADIATION DETECTION SIGNAL PROCESSING CIRCUIT

If a crystal detector 142 fabricated of thallium-activated sodium iodide is utilized in place of the preferred solid-state detector then the radiation detection signal processing circuit configuration shown in FIG. 6 is preferably utilized for each of the detection channels in place of the detection signal processing circuit 67 configuration previously described with reference to FIG. 3.

Referring now to FIG. 6, the photomultiplier tube 146 for a typical detection channel in which a thallium-activated sodium iodide crystal is utilized as the detector 142 is preferably a conventional 10-stage photomultiplier having an individual voltage dividing network, such as is provided by an RCA type 7767 multiplier phototube which has ten electrostatically focused dynode stages with in-line arrangement thereof. In the system 10 of the present invention, such a tube 146 is preferably magnetically shielded. A conventional high voltage power supply 170 is connected to the photomultiplier tube 146 via a conventional gain adjustment potentiometer 172 having an adjustable wiper arm 173 so as to supply the required voltage necessary to excite the photo-cathode to its range of sensitivity to light. If desired, a single high voltage power supply 170 may be utilized to supply power to any number of photomultiplier tubes 146 in the detector assembly array. In such instance, a different gain adjustment potentiometer 172 is preferably provided for each tube 146 so as to properly adjust the voltage for each of the individual detectors 142. The output of the photomultiplier tube 146 is connected through a conventional pre-amplifier 174 and amplifier 176 to a conventional discriminator stage or variable pulse height detector stage 178 whose function is to eliminate all pulses or voltages below and above a specified level so as to eliminate low voltage electrical noise and background counts or emissions outside of the energies to be detected. In addition the discriminator stage 178 preferably is provided with conventional means 179, such as a variable potentiometer, for shifting the discriminator to another threshold level which would be appropriate for a different radioisotopic gas as well as to the optimal levels for the preferred radioisotopic gas Xenon¹³³. The output of the discriminator stage 178 is preferably fed through a conventional summing gate and pulse stretcher network 180 whose output is preferably fed to the general pur-

pose computer 61 via the minicomputer buffer 60, if desired.

LOCAL CEREBRAL BLOOD FLOW DETERMINATION METHOD

General Description

Now generally describing the preferred method of the present invention in accordance with which the local cerebral blood flow determination system 10 previously described is preferably utilized. In any local cerebral blood flow determination there are significant physical factors which affect this determination but which may be considered to be constants for a given patient whose local cerebral blood flow is being determined. However, these significant constants must preferably be initially determined for the patient. In utilizing the system 10 in accordance with the preferred method of the present invention, the local cerebral blood flow is ultimately determined in accordance with the well known Fick-Kety expression of which one form is

$$\int_{m_1}^{m_2} C_i^2 W dt = F \int_{m_1}^{m_2} (C_a - C_v) dt$$

where C_i equals the brain tissue radioisotope concentration preferably in microcuries per gram; $(C_a - C_v)$ equals the arterialvenous blood radioisotope concentration difference, preferably in microcuries per gram; m_1 equals the starting time of the local cerebral blood flow determination, preferably in seconds, F equals the local cerebral blood flow, preferably in microliters per gram per minute; W equals the weight of the tissue comprising the detection zone for which the local cerebral blood flow is being determined; and m_2 equals the number of seconds after the beginning of the first inhalation by the patient during the data collection period.

As will be explained in greater detail hereinafter, this Fick-Kety expression is preferably utilized for each of the detection zones 81 in which it is desired to determine the local cerebral blood flow therefor. Furthermore, it should be noted that the local cerebral blood flow determined in accordance with the preferred method of the present invention is preferably a quantitative determination of cerebral blood flow for that detection zone 81 for which the Fick-Kety expression is utilized (as opposed to a mean rate of flow for a region) which, as will be explained in greater detail hereinafter, is possible because the various factors in the Fick-Kety expression can be localized for each detection zone 81 in accordance with the preferred method of the present invention. C_v and C_i are related by the expression C_v (venous concentration) = C_i (brain tissue concentration) $\div \tau$ (tissue/blood partition coefficient for the gas), and the value of τ can be determined. Furthermore, since the specific gravity of brain tissue is approximately equal to one, the weight of the tissue W is approximately equal to the volume of the detection zone 81, which is also a known factor as it is predetermined by the volume of the model which defines each of the detection zones 81. Therefore, the only unknowns involved in applying this Fick-Kety expression to the determination of a local cerebral blood flow is the tissue radiation concentration quantity C_i and the arterial blood radiation concentration quantity C_a . It is in de-

termining these quantities for a preselected discrete detection zone **81** that the preferred method of the present invention is most useful.

DETAILED DESCRIPTION

Now describing the preferred method of the present invention in greater detail. In utilizing the system **10** of the present invention in accordance with the preferred method of the present invention, the headgear apparatus **66** is preferably aligned on the patient's head so as to provide a specific and reproducible relationship to the patient's brain. As was previously mentioned, the headgear apparatus **66** is preferably aligned so as to be oriented as shown in FIG. 5 with the glabellar-sellar plane being the basal plane for the detector array of the rectangular helmet **82**.

When the headgear apparatus **66** is so aligned, the other two preferred reference planes of the three reference planes which define the orientation or alignment of the headgear apparatus **66** are, as was previously mentioned, the mid-sagittal plane and the plane perpendicular to the basal plane and the mid-sagittal plane and tangent to the forehead, and the detection zones **81** of the patient's brain are defined as shown in FIG. 5 by the dotted lines.

As was previously mentioned, the length of air space between the associated detector assembly and the portion of the patient's scalp opposite that detector assembly affects the radiation emission detected thereby. Therefore, it is both preferable and desirable that the length of air space between a particular detector assembly and the skull of the patient be known for purposes of the local cerebral blood flow determination. Accordingly, if a separate head-gauge is utilized to measure the lengths of these air spaces, as was previously mentioned, the gauges **130** through **138**, inclusive, located in the helmet housing **82** must be checked to insure that the patient's head retain exactly the same position during the actual data collection period as during the time when the air space measurements were made.

In addition, compensation should also preferably be made for the thickness of the patient's skull and the degree of bone mineralization as these factors may introduce errors in the measurements of radiation emission by the P and Q detector assemblies. These factors may be determined by ultrasonic techniques such as are employed for neurologic echo-scanning. However, if the errors introduced by variations in bone mineralization are tolerable, then no additional compensation need be made for this factor. If the errors introduced by variations in skull thickness can be tolerated, no additional compensation need be made for this factor either. Corrections for mean skull thickness and density are then sufficient.

Once the headgear apparatus **66** is aligned in the preferred predetermined orientation or alignment position, the balance of the local cerebral blood flow determination may be performed in accordance with the preferred method of the present invention. The radioactive gas, which preferably has at least two gamma energy levels, such as Xenon¹³³ preferably, although other gases such as Krypton 85 or gaseous mixtures of two radioactive gases each having a different gamma energy level may be utilized if desired, is then dissolved in the patient's blood by means of the closed inhalation-exhalation breathing system associated with the

mask **12**. Throughout the actual data collection period, the patient receives his air supply through this closed inhalation-exhalation breathing system. When the actual data collection is to commence, that is preferably when the patient's breathing conditions, such as oxygen and carbon dioxide pressure, are stabilized as indicated by the carbon dioxide and oxygen analyzers **76** and **77**, the minicomputer buffer **60** preferably sends a signal to the end-tidal sampler **62** which activates the sampler **62** to turn it ON so as to preferably synchronize the initiation of end-tidal sampling and the resultant radiation emission detection or counting times in the Xenon¹³³ gas analyzer detector assembly **65** with that of the P and Q detector assemblies associated with the headgear apparatus **66** detector array. After this initial activation, the end-tidal sampler **62** functions in a conventional manner throughout the duration of the data collection period, and its operation will not be described in greater detail hereinafter. Suffice it to say that the end-tidal sampler **62** relay mechanism (not shown) is actuated at the beginning and end of each breath by airway pressure changes within the closed breathing system resulting from these conditions; the end-tidal sampler **62** functioning to extract end-tidal samples when activated by the gas pressure within the mask **12** reaching a given pressure at the end of the breath and continuing to sample until negative pressure arises at the beginning of the next breath.

Subsequent to the initial turning ON of the endtidal sampler **62**, signifying the beginning of the actual data collection period, the actuation of the end-tidal sampler **62** relay at the end of the patient's next breath actuates the Xenon¹³³ gas injector **40** solenoids (not shown) associated with valves **42** and **50**. Gas injector **40** thereafter injects a predetermined amount of the radioisotopic Xenon¹³³ gas into both the inhalation and the exhalation chambers **28** and **30**, respectively, through the respective intermittently opened valves **42** and **50**, which thereafter close, and associated conduits **44** and **48**, respectively. The injected radioisotopic gas mixes with the air already contained in the chambers **28** and **30**.

As the patient continues to breath normally, the airway pressure in inhalation conduit **16**, and the closed breathing system as a whole, changes and demand valve **20** opens at the beginning of the next breath. The end-tidal sampler **62** relay is again actuated in response to this initial pressure change, which preferably sends an initial signal to the minicomputer **60**, or if desired, directly to the measuring devices; namely, the radiation detection signal processing circuits **67** and **74**, placing them in the data collection mode. This signal is preferably only sent at this initial time, the circuits **67** and **74** remaining in this mode throughout the balance of the data collection period. In addition, a signal is also sent to the gas purification unit **32** which signal activates the unit **32** so as to refill the inhalation chamber **28** from the exhalation chamber **30** via the gas purification unit **32** which removes excess carbon dioxide, adds oxygen to the closed breathing system, and filters the air contained thereon.

Each subsequent breath of the patient during the data collection period causes the actuation of the endtidal sampler **62** relay mechanism at the beginning of the breath; the resultant operation of gas injector **40** to inject a predetermined quantity of radioisotopic Xenon¹³³ gas only into the exhalation chamber **30**,

valve 50 preferably remaining closed throughout the balance of the data collection period; the actuation again of the end-tidal sampler 62 relay mechanism at the end of the breath; and the resultant operation of gas purification unit 32 to return the injected gas well mixed to the inhalation chamber 28 to be breathed by the patient. In this manner, the radioisotopic gas concentration in the blood is caused to rise steeply since each breath contains a still higher concentration of the radioisotopic gas.

Arterial Blood Radiation Concentration Determination

Preferably, as was previously mentioned, in order to determine the arterial blood radiation concentration quantity C_a in accordance with the preferred method of the present invention, the alveolar or end-tidal gas concentration at the end of each breath is sampled in the manner previously described above. This end-tidal gas sample is drawn through the coiled passageway 64 which, as was previously mentioned, contains a known total volume of the sampled gas. Detector assembly 65 "looks" at the coiled passageway 64 and preferably measures the gamma radiation emission emanating therefrom, which is equivalent to the gamma radiation emission in that known sample volume. The gamma radiation emission entering collimator 140 which is incident on detector 142 is absorbed and causes the generation of secondary electrons which are amplified in the radiation detection signal processing circuit 74 to provide a pulse or signal which is the equivalent of the resultant radiation emission count of detector assembly 65. Since the lung gas in contact with the blood has essentially the same partial pressure as the gas contained in arterial blood, and since it is this equilibrated gas which appears in the last part of each breath, the gas concentration in the last part of each breath is essentially equivalent to the concentration of the radioisotopic gas in arterial blood. This signal, which therefore is representative of the arterial blood radiation concentration, is then preferably fed to the minicomputer buffer 60 for temporary storage before being passed to the general purpose computer 61 for processing along with the other collected data.

As will be explained in greater detail hereinafter, each time an end-tidal sample is taken and the resultant arterial blood radiation concentration signal produced in accordance therewith, a point on an arterial concentration curve is "created." Most preferably, the data collection period duration is one minute and six determinations at 10 second intervals m_0 , and m_1, m_2, m_3, m_4 , and m_5 , as shown in FIG. 4, are made so as to "construct" an arterial concentration curve for purposes of the local cerebral blood flow determination in accordance with the Fick-Kety expression, this arterial blood radiation concentration quantity being represented by the term C_a .

BRAIN TISSUE RADIATION CONCENTRATION DETERMINATION

Now referring to the operation of the system 10 in accordance with the preferred method of the present invention so as to provide the brain tissue radioisotope concentration quantity C_t for, preferably, each of the detection zones 81 so that this data may be processed in accordance with the previously stated Fick-Kety expression. The radioactive gas, which as was previously mentioned dissolves in the blood passing through the

lungs, is delivered to the brain parts or detection zones 81 in proportion to the local rate of blood flow. The resultant gamma radiation emissions detected by the various P and Q detector assemblies included in the head-gear apparatus 66 detector array are a function of the amount of radioactive gas deposited, or radioisotope concentration, in the various detection zones 81 and the effective field of view associated with a given detector assembly.

As was previously described, in the preferred method of the present invention, the 144 blocks or zones 81 which describe the brain volume are preferably subdivided into four equal 36 zone quadrants of $3 \times 3 \times 4$ blocks or zones 81 each which are each preferably similarly solved for the valves of the radioisotope concentration in each of the zones 81 by solving four sets of 26 linear simultaneous equations, one set for each quadrant, yielding the radioisotope concentrations for each of the 144 detection zones 81. In the alternative, if desired, the brain volume can be looked at in its entirety as a whole and 144 simultaneous equations for the radioisotope concentrations in the 144 various zones 81 be solved as a group. The P and Q detector assemblies "look" at each quadrant of the brain from 36 different positions (or locations), for a total of 144 different positions (or locations), and the degree of resolution or localization observed by each P and Q detector assembly, that is, its effective field of view, is determined by the collimator 140 associated with the particular P or Q detector 142. In order to properly determine the rate of local cerebral blood flow, the radioactive gas must be administered to the patient in sufficiently high concentration to provide a statistically acceptable counting rate, or quantity of detectable radiation emission, in the preferred sampling interval which is 10 seconds, but not so high that the patient is endangered by the radiation. As was previously mentioned, the preferred square lumen collimator 140 has a length which provides the preferred substantially three detection zone 81 useful depth of view for the preferred effective field of view for a P group detector and two detection zone 81 useful depth of view for a Q group detector with a proportionate amount of such zones 81 which are in a three-by-three matrix array or two-by-two matrix array with the zone at the point of "entry" into the volume for the angled P and Q groups of detectors, respectively. This preferred collimator 140, which preferably differs in length for the p and Q groups of detectors, respectively, represents the optimal compromise between an acceptable level of radioactive gas isotope administration and the collimator size suitable for localization. Each preferred P or Q detector 142, therefore, sees its associated preferred effective field of view of either three or two detection zones in depth, respectively, the total or resultant radiation emission observed by each detector 142, as was previously mentioned, being due to the radioisotope concentration present in the portions of each of the detection zones 81 contained within the effective field of view of the particular detector 142.

This relationship between the resultant radiation emission detected or observed by a given detector 142 and the radioisotope concentration present in the portions of each of the detection zones 81 contained within the preferred effective field of view of the detector 142 may be preferably defined by the following expressions in which we have assumed, for purposes of

explanation, that all the detection zones **81** are included within the bone surrounding the patient's brain, although it would be obvious to one of ordinary skill how to readily modify the following expressions to take into account the surrounding bone if the detection zones were defined so as to include this bone which, in such instance, could be considered as non-radioactive source bearing material. Before defining these expressions, however, a nomenclature for the detection zones **81** including the effective field of view, preferably must first be established.

For purposes of explanation, the 36 detection zones for the typical quadrant illustrated in FIGS. 9-23, which is the front-right quadrant, are labeled Z-1 through Z-36, inclusive, with the uppermost horizontal plane 3x3 matrix array labeled Z-1 through Z-3 in the outermost row, Z-4 through Z-6 in the next adjacent row and Z-7 through Z-9 in the innermost row. Similarly the next adjacent horizontal plane matrix array is labeled, Z-10 through Z-12 in the outermost row, Z-13 through Z-15 and Z-16 through Z-18; the array in the plane below being labeled Z-19 through Z-21, Z-22 through Z-24, and Z-25 through Z-27, respectively; and the array in the lowermost plane being labeled Z-28 through Z-30, Z-31 through Z-33 and Z-34 through Z-36, respectively. The P and Q detectors are each labeled with a subscript containing the number of the zone **81** through which the brain is "looked" into or "entered" and a letter indicating the direction of "entry", "G" indicating entry from the top of the 36 zone array, "C" indicating entry from the right side, "F" indicating entry on a preferably 45° angle from the right side above towards the left side below, "A" indicating entry from the front, "H" indicating entry on a preferably 45° angle from the front above towards the rear below and "B" indicating entry on a preferably 45° angle in the horizontal plane from the right towards the left.

For purposes of explanation, based upon the preferred effective fields of view, and the preferred planes of orientation of the headgear apparatus **66**, the given P or Q detector assembly, for which the expressions are defined, shall be assumed to be located, in the most general case, adjacent the center of the outermost surface of the zone **81** through which the appropriate P or Q detector assembly looks into the brain volume, those P and Q detector assembly which look into the brain volume at an angle generally being assumed to be located adjacent the center of the line of intersection of the planes of the two normal adjacent outermost surfaces of the zone **81** through which the appropriate angled P or Q detector assembly looks into the brain volume. Furthermore, as was previously mentioned, the

effective field of view of each P or Q detector assembly is preferably symmetrical about the axis of view, the field of view including a proportionate amount of each zone which can be "seen". For purpose of explanation, the proportionate amounts of the zones **81** included within the effective field of view of a Q detector assembly are generally designated " $\bar{m}$ " for the first zone in depth and " $\bar{n}$ " for the second zone in depth for the non-angled detectors, and " $\bar{p}$ " for the first zone in depth and " $\bar{q}$ " and " $\bar{r}$ " for the portions of the second zones in depth for the angled detectors. Similarly, the proportionate amounts of the zones **81** included within the effective field of view of a P detector assembly are generally designated " $\bar{a}$ " for the first zone in depth, " $\bar{b}$ " for the second zone in depth and " $\bar{c}$ " for the third zone in depth for the non-angled detectors, and " $\bar{d}$ " for the first zone in depth, " $\bar{e}$ " and " $\bar{f}$ " for portions of the second zones in depth and " $\bar{g}$ " and " $\bar{h}$ " for portions of the third zones in depth for the angled detectors. Furthermore, for certain of both the P and Q detector assemblies, a proportionate amount of an additional portion of another zone within the zone adjacent the first zone in depth within the effective field of view is generally designated " $\bar{k}$ ". The general letters " $\bar{a}$ ", " $\bar{b}$ ", " $\bar{c}$ ", " $\bar{d}$ ", " $\bar{e}$ ", " $\bar{f}$ ", " $\bar{g}$ ", " $\bar{h}$ ", " $\bar{k}$ ", " $\bar{m}$ ", " $\bar{n}$ ", " $\bar{p}$ ", " $\bar{q}$ " and " $\bar{r}$ " are factors for each appropriate zone **81** related to the efficiency " ϵ " of the detection system; the fraction " δ " of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination (loss due to radioactive decay); the factor " s " expressing the attenuation of radiation by the scalp and skull; the area " A_d " of the detector; the dimensional constant " K "; the total attenuation constant " u " for gamma radiation of a specific energy (in square centimeters/gram); the total thickness " t " of brain tissue through which the radiation is attenuated (in grams/square centimeter); the distance " l " from the center of the particular zone to the detector along the central axis of view; the volume of the zone " V_T "; the collimator factor " f_c " for a given zone, which is the projected area of the detector divided by the total area of the detector as seen through the collimator; VP zone or VQ zone represents the integration limit of the total volume of the particular P or Q zone; and "I zone", representing the radiation concentration (in microcuries) of radioisotope contained within a given zone; "P subscript" and "Q subscript" representing the detected resultant radiation emission or counting rate resulting from the contributions of all the detection zones **81** within the effective field of view of the particular P or Q detector assembly, where the following general expressions define these general letters as follows with the term C being a constant equivalent to the expression

$$\frac{KA_d \epsilon \delta s}{V_T} :$$

$$\bar{a} = C \int_0^{V_{1PZONE}} \frac{e^{-\mu t_{1PZONE}}}{l_{1PZONE}^2} f_{c1PZONE} dV$$

$$\bar{b} = C \int_0^{V_{2PZONE}} \frac{e^{-\mu t_{2PZONE}}}{l_{2PZONE}^2} f_{c2PZONE} dV$$

$$\bar{c} = C \int_0^{V_{3PZONE}} \frac{e^{-\mu t_{3PZONE}}}{l_{3PZONE}^2} f_{c3PZONE} dV$$

$$\bar{d} = C \int_0^{V_{1'PZONE}} \frac{e^{-\mu t_{1'PZONE}}}{l_{1'PZONE}^2} f_{d1'PZONE} dV$$

$$\bar{e} = C \int_0^{V_{2'PZONE}} \frac{e^{-\mu t_{2'PZONE}}}{l_{2'PZONE}^2} f_{e2'PZONE} dV$$

$$\bar{f} = C \int_0^{V_{2''PZONE}} \frac{e^{-\mu t_{2''PZONE}}}{l_{2''PZONE}^2} f_{f2''PZONE} dV$$

$$\bar{g} = C \int_0^{V_{3'PZONE}} \frac{e^{-\mu t_{3'PZONE}}}{l_{3'PZONE}^2} f_{g3'PZONE} dV$$

$$\bar{h} = C \int_0^{V_{3''PZONE}} \frac{e^{-\mu t_{3''PZONE}}}{l_{3''PZONE}^2} f_{h3''PZONE} dV$$

$$\bar{k} = C \int_0^{V_{1'''P/QZONE}} \frac{e^{-\mu t_{1'''P/QZONE}}}{l_{1'''P/QZONE}^2} f_{k1'''P/QZONE} dV$$

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^2} f_{m1QZONE} dV$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^2} f_{n2QZONE} dV$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^2} f_{p1'QZONE} dV$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^2} f_{q2'QZONE} dV$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^2} f_{r2''QZONE} dV$$

where the zonal subscripts "1 Q zone", "2 Q zone", and "1 P zone", "2 P zone" and "3 P zone" refer to the particular first (1), second (2) or third (3) zone in depth for the non-angled appropriate Q and P detectors, respectively; "1' Q zone", "2' Q zone" and "2'' Q zone", and "1' P zone", "2' P zone", "2'' P zone", "3' P zone" and "3'' P zone" refer to the particular appropriate portions of the first (1'), second (2' and 2'') and third (3' and 3'') zone in depth for the angled Q and P detectors, respectively; and "1''' P/Q zone" for the portion of the first (1''') zone in depth for the appropriate angled P or Q detector.

Utilizing the general letters defined above, which relate to the geometric proportionate amount of the various detection zones contained within the effective field of view of the appropriate detector, taken together with the previously mentioned nomenclature the 36 simultaneous equations representative of the 36 detection zones within a typical quadrant can be written as follows, taking the quadrant illustrated in FIGS. 9-23 by way of example:

Referring to FIGS. 9A-9E:

$$Q_{G4} = \bar{m}I_4 + \bar{n}I_{13}$$

$$Q_{C1} = \bar{m}I_1 + \bar{n}I_4$$

$$Q_{F1} = \bar{p}I_1 + \bar{q}I_4 + \bar{q}I_{10} + \bar{r}I_{13}$$

$$Q_{C10} = \bar{m}I_{10} + \bar{n}I_{13}$$

Referring to FIGS. 10A-10E:

$$Q_{G2} = \bar{m}I_2 + \bar{n}I_{11}$$

$$Q_{G3} = \bar{m}I_3 + \bar{n}I_{12}$$

$$Q_{A12} = \bar{m}I_{12} + \bar{n}I_{11}$$

$$Q_{H3} = \bar{p}I_3 + \bar{q}I_2 + \bar{q}I_{12} + \bar{r}I_{11}$$

Referring to FIGS. 11A-11E:

$$Q_{G8} = \bar{m}I_8 + \bar{n}I_{17}$$

$$Q_{G9} = \bar{m}I_9 + \bar{n}I_{18}$$

$$Q_{A18} = \bar{m}I_{18} + \bar{n}I_{17}$$

$$Q_{H9} = \bar{p}I_9 + \bar{q}I_8 + \bar{q}I_{18} + \bar{r}I_{17}$$

Referring to FIGS. 12A-12D:

$$Q_{G5} = \bar{m}I_5 + \bar{n}I_{14}$$

$$Q_{G6} = \bar{m}I_6 + \bar{n}I_{15}$$

$$Q_{A15} = \bar{m}I_{15} + \bar{n}I_{14}$$

$$Q_{H6} = \bar{p}I_6 + \bar{q}I_5 + \bar{q}I_{15} + \bar{r}I_{14}$$

Referring to FIGS. 13A-13E:

$$Q_{A24} = \bar{m}I_{24} + \bar{n}I_{23}$$

$$Q_{A21} = \bar{m}I_{21} + \bar{n}I_{20}$$

$$Q_{B21} = \bar{p}I_{21} + \bar{q}I_{20} + \bar{q}I_{24} + \bar{r}I_{23}$$

$$Q_{C20} = \bar{m}I_{20} + \bar{n}I_{23}$$

Referring to FIGS. 14A-14D:

$$Q_{A33} = \bar{m}I_{33} + \bar{n}I_{32}$$

$$Q_{A30} = \bar{m}I_{30} + \bar{n}I_{29}$$

$$Q_{B30} = \bar{p}I_{30} + \bar{q}I_{29} + \bar{q}I_{33} + \bar{r}I_{32}$$

$$Q_{C29} = \bar{m}I_{29} + \bar{n}I_{32}$$

Referring to FIGS. 15A-15C and 16A and 16B:

$$P_{A9} = \bar{a}I_9 + \bar{b}I_8 + \bar{c}I_7$$

$$P_{A27} = \bar{a}I_{27} + \bar{b}I_{25} + \bar{c}I_{25}$$

Referring to FIGS. 17A-17C:

$$P_{C21} = \bar{a}I_{21} + \bar{b}I_{24} + \bar{c}I_{27}$$

Referring to FIGS. 18A, 18B, 19A, 19B and 20A-20C:

$$P_{B12} = \bar{d}I_{12} + \bar{e}I_{11} + \bar{e}I_{15} + \bar{f}I_{14} + \bar{g}I_{13} + \bar{g}I_{17} + \bar{h}I_{16}$$

$$Q_{C19} = \bar{m}I_{19} + \bar{n}I_{22}$$

$$P_{A36} = \bar{a}I_{36} + \bar{b}I_{35} + \bar{c}I_{34}$$

$$Q_{B29} = \bar{k}I_{30} + \bar{p}I_{29} + \bar{q}I_{28} + \bar{q}I_{32} + \bar{r}I_{31}$$

Referring to FIGS. 21A, 21B, 22A, 22B, 23A and 23B:

$$P_{G1} = \bar{a}I_1 + \bar{b}I_{10} + \bar{c}I_{19}$$

$$P_{G7} = \bar{a}I_7 + \bar{b}I_{16} + \bar{c}I_{25}$$

$$Q_{C28} = \bar{m}I_{28} + \bar{n}I_{31}$$

$$P_{F11} = \bar{k}I_{12} + \bar{d}I_{11} + \bar{e}I_{14} + \bar{e}I_{20} + \bar{f}I_{23} + \bar{g}I_{26} + \bar{g}I_{32} + \bar{h}I_{35}$$

$$P_{F12} = \bar{k}I_{13} + \bar{d}I_{12} + \bar{e}I_{15} + \bar{e}I_{21} + \bar{f}I_{24} + \bar{g}I_{27} + \bar{g}I_{33} + \bar{h}I_{36}$$

The above expressions preferably include corrections for tissue attenuation of the gamma radiation originating at various depths within the brain; i.e., in other detection zones, as well as for the inverse square relationship between counts detected and distance.

By way of example, substituting the equivalent expressions for $\bar{a}$, $\bar{b}$, $\bar{c}$, $\bar{d}$, $\bar{e}$, $\bar{f}$, $\bar{g}$, $\bar{h}$, $\bar{k}$, $\bar{m}$, $\bar{n}$, $\bar{p}$, $\bar{q}$ and $\bar{r}$, in

the equations for Q_{G4} , Q_{B29} , P_{A9} and P_{F11} can be rewritten as follows, such substitutions in the balance of the above equations representative of the detection zones being obvious for the remaining equations for the p and Q detector; substituting the appropriate zones for the "zone" subscripts of the various expressions by inspection with respect to FIGS. 9-23:

$$Q_{G4} = \frac{KA_d \epsilon \delta s}{V_T} \left[\int_0^{V_{z-4}} \frac{e^{-\mu t_{z-4}}}{l_{z-4}^2} f_{cz-4} dVI_4 + \int_0^{V_{z-13}} \frac{e^{-\mu t_{z-13}}}{l_{z-13}^2} f_{cz-13} dVI_{13} \right]$$

$$Q_{B29} = \frac{KA_d \epsilon \delta s}{V_T} \left[\int_0^{V_{z-30}} \frac{e^{-\mu t_{z-30}}}{l_{z-30}^2} f_{cz-30} dVI_{30} + \int_0^{V_{z-29}} \frac{e^{-\mu t_{z-29}}}{l_{z-29}^2} f_{cz-29} dVI_{29} \right. \\ \left. + \int_0^{V_{z-28}} \frac{e^{-\mu t_{z-28}}}{l_{z-28}^2} f_{cz-28} dVI_{28} + \int_0^{V_{z-32}} \frac{e^{-\mu t_{z-32}}}{l_{z-32}^2} f_{cz-32} dVI_{32} \right. \\ \left. + \int_0^{V_{z-31}} \frac{e^{-\mu t_{z-31}}}{l_{z-31}^2} f_{cz-31} dVI_{31} \right]$$

$$P_{A9} = \frac{KA_d \epsilon \delta s}{V_T} \left[\int_0^{V_{z-9}} \frac{e^{-\mu t_{z-9}}}{l_{z-9}^2} f_{cz-9} dVI_9 + \int_0^{V_{z-8}} \frac{e^{-\mu t_{z-8}}}{l_{z-8}^2} f_{cz-8} dVI_8 \right. \\ \left. + \int_0^{V_{z-7}} \frac{e^{-\mu t_{z-7}}}{l_{z-7}^2} f_{cz-7} dVI_7 \right]$$

$$P_{F11} = \frac{KA_d \epsilon \delta s}{V_T} \left[\int_0^{V_{z-2}} \frac{e^{-\mu t_{z-2}}}{l_{z-2}^2} f_{cz-2} dVI_2 + \int_0^{V_{z-11}} \frac{e^{-\mu t_{z-11}}}{l_{z-11}^2} f_{cz-11} dVI_{11} \right. \\ \left. + \int_0^{V_{z-14}} \frac{e^{-\mu t_{z-14}}}{l_{z-14}^2} f_{cz-14} dVI_{14} + \int_0^{V_{z-20}} \frac{e^{-\mu t_{z-20}}}{l_{z-20}^2} f_{cz-20} dVI_{20} \right. \\ \left. + \int_0^{V_{z-23}} \frac{e^{-\mu t_{z-23}}}{l_{z-23}^2} f_{cz-23} dVI_{23} + \int_0^{V_{z-26}} \frac{e^{-\mu t_{z-26}}}{l_{z-26}^2} f_{cz-26} dVI_{26} \right. \\ \left. + \int_0^{V_{z-32}} \frac{e^{-\mu t_{z-32}}}{l_{z-32}^2} f_{cz-32} dVI_{32} + \int_0^{V_{z-35}} \frac{e^{-\mu t_{z-35}}}{l_{z-35}^2} f_{cz-35} dVI_{35} \right]$$

can then readily be rewritten by inspection by anyone with ordinary mathematical skill for any of the 144 total zones **81** within any of the four quadrants of the brain volume, understanding, of course, that the directions of view of the various P and Q detector assemblies for each of the quadrants is from the outside "looking" in, only the subscripts relating to the various zones being altered in accordance with its given designation of Z-1 through Z-144 depending on the quadrant, one quadrant being Z-1 through Z-36 (illustrated in FIGS. 9-23), another Z-37 through Z-72, another Z-73 through Z-108 and the other Z-109 through Z-144. Thus, the 144 counting expressions for the various parts of the same tissue volume can readily be provided either quadrant by quadrant or in its entirety. Since the quantity of detected radiation emission is directly related to the tissue radioisotope concentrations in the portions of the detection zones contained within the associated effective field of view, the tissue radioisotope concentration quantity " C_i ", which is equivalent to "I" for a given detection zone, for each of the individual detection zones can be determined from the actual resultant radiation emissions observed by the detector array of the helmet housing **82**, by preferably solving these 36 counting equations or expressions for each quadrant simultaneously, or, if desired, by solving the resultant 144 expressions for the entire volume simultaneously.

As was previously mentioned, the minicomputer buffer **60** preferably collects the counted radiation emissions which have been processed by the radiation detection signal processing circuitry **67**, preferably completing a full-scan cycle for all of the detectors in the array before the last pulse or signal from the array has decayed, so as to provide tissue radiation emission signals from each of the 144 detectors in the four quadrants associated with the helmet housing **82** every 10 seconds for the preferred one minute data collection period during which the blood radioisotope concentration is raised by breathing the Xenon **133** gas. At the end of the preferred one minute data collection period, the minicomputer buffer **60** then preferably transfers this data to the general purpose computer **61**, if it is available, which is programmed in conventional fashion to solve the four sets of 36 simultaneous equations each (144 simultaneous equations in total) for each 10-second period. The minicomputer **60**, then preferably goes "OFF-line" and this "computer-OFF" signal is utilized to operate valves **52** and **54** so as to change the mask **12** intake from the inhalation chamber **28** to room air, and the mask **12** gas outflow to the exhaust hood instead of the exhalation chamber **30**. Thereafter in a few breaths, most of the radiosiotopic gas will have been breathed out of the patient and into the exhaust hood. If desired, if the general purpose computer **61** is not available at this time, the minicomputer **60** may go "OFF-line" so as to operate valves **52** and **54** and store the collected data for processing at a later time by the general purpose computer **61**. In any event, if desired, at the completion of the data collection period, when the minicomputer **60** has collected all the data, it may transmit a signal which can be utilized to operate valves **52** and **54**.

After the collected resultant radiation emission data has been transferred to the general purpose computer **61**, the computer **61** then preferably "constructs" a curve of the rate of change in tissue radioisotope concentration "I" or C_i in each brain part or detection zone, utilizing the six tissue radioisotope concentra-

tions "I" quantified for each brain part or detection zone at the 10-second sampling intervals. Such a representative curve is shown by way of example in FIG. 4 by curve **210**, the sampling intervals being represented by m_0, m_1, m_2, m_3, m_4 , and m_5 . As was previously mentioned, the computer **61** also "constructs" a curve for the rate of change of radioisotope concentration C_a in arterial blood for these time intervals m_0 through m_5 , inclusive, which curve is represented by the curve **212** in FIG. 4. Curve **210** provides the tissue radioisotope concentration characteristic C_i for the detection zone while curve **202** provides the arterial blood radioisotope concentration characteristic C_a . The computer **61** thereafter preferably determines the rate of local cerebral blood flow for each of the 144 detection zones or brain parts utilizing these characteristics or curves **210** and **212** in accordance with the previously described Fick-Kety expression, the shaded area in FIG. 4 between curves **210** and **212** illustratively representing the integral portion of this Fick-Kety expression.

When utilizing the system of the present invention, if desired, the computer **61** may be programmed in a conventional manner to determine cerebral blood flow differences above and below a normal level and only to provide a print-out of the data within specific levels of abnormality, such as a print-out of all detection zones in which the blood flow is 20 to 40 percent below normal. Furthermore, if desired, the computer **61** may be programmed in a conventional manner to provide a printout of the local cerebral blood flow in each of the detection zones irrespective of whether such blood flow is normal or abnormal or, if desired, the computer **60** may be programmed in a conventional manner to write the local cerebral blood flows on a "brain map". It will become apparent to one of ordinary skill in the art, that the local cerebral blood flow data may be utilized in a multiplicity of fashions depending on the desired diagnostic use of this information.

Summarizing the preferred method of the present invention, the headgear apparatus is aligned in the predetermined position on the head of the patient; the patient then inhales the radioactive gas in a closed inhalation-exhalation breathing system; the last part of the patient's breath is end-tidal sampled at 10-second intervals during a one minute data collection period to provide an arterial blood radioisotope concentration characteristic; the headgear apparatus detector array, which "looks" at the subdivided 144 detection zones in the brain in four quadrants of 36 zones each, quantifies the resultant radiation emission at each detector at the 10-second sampling intervals, which information is processed in accordance with the appropriate counting expression P or Q for each detection zone in each of the quadrants to provide four sets of 36 simultaneous equations each (144 in total) to the computer which solves these equations to provide a tissue radioisotope concentration characteristic; and the computer processes the tissue radioisotope concentration characteristic and the arterial blood radioisotope concentration characteristic in accordance with the Fick-Kety expression to provide a local cerebral blood flow for one or more of the 144 detection zones.

Alternative Method

Referring now to FIGS. 2A, 2B, 7A, 7B and 8 an alternative method for determining discrete localized blood flow in a large volume of tissue, such as the brain shall now be described, utilizing a determination of local cerebral blood flow within the brain volume, by

way of example, so as to be consistent with the description of the preferred method of the present invention. This alternative method differs from the previously described preferred method, as will be described in greater detail hereinafter, in that only a single gamma energy level is detected, such as the gamma radiation 81 Kev energy level radiation of the preferred Xenon¹³³ gas, all detectors 95 detected the same energy level, and the brain volume is "looked" at as a whole rather than by quadrants to yield the 144 simultaneous linear equations defining the radioisotope concentration within the various zones 81 describing the brain volume. The detector assembly 95 and electronics are preferably substantially structurally and functionally identical with that previously described as is the balance of the apparatus utilized in practicing the method of the present invention, the alternative method primarily differing in the tissue of determining the brain tissue radiation concentration, although the collected data is preferably processed in accordance with the aforementioned Fick-Kety expression. For purposes of clarity, however, the relevant portions of the apparatus preferably utilized in practicing the alternative method of the present invention shall be reiterated where it is deemed necessary.

Referring now to FIGS. 2A and 2B, in order to define the 144 detection zones 81 and the brain, which are the radiation emission detection zones for purposes of the local cerebral blood flow determination, the headgear apparatus 66, as was previously mentioned, preferably includes the substantially rectangular hollow helmet housing 82 having front portion 84, rear portion 86, top portion 88, side portions 90 and 92, and bottom portion 94 through which the head of the patient is inserted. Preferably, a plurality of substantially identical individual detector assemblies 95, equal in number to the plurality of radiation emission detection zones, 81, which, as was previously mentioned, in the preferred embodiment is 144 are arrayed in parallel and angular 83 modular banks on the front 84, rear 86, top 88, and sides 90 and 92 of the helmet housing 82, along edges, and in a pair of modular banks 96 and 98 of these individual detector assemblies 95 as required to yield independent linear equations. Each of the detector assemblies 95 (which are preferably functionally and structurally substantially identical to the P and Q detectors except differing in their associated effective field of view) is preferably arrayed along an axis normal to the plane in which the respective associated modular bank 83, 84, 86, 88, 90, 92, 96 and 98 lies. Modular banks 96 and 98 are preferably angularly related to the side portions of the helmet housing, respectively, at an angle of 135 degrees with the cube diagonals. For purposes of illustration, modular banks 96 and 98 are shown rotated 90 degrees in to the plane of view with respect to their actual preferred positions. One of the detector assemblies 95a and 95b is shown in cross-section in FIGS. 2a and 2b, respectively, for purposes of explanation, the field of view of assemblies 95a and 95b being preferably identical in volume.

As shown in FIG. 2B, the top portion 88 of the helmet housing 82 preferably includes six modular banks or columns of detector assemblies 95, columns 102, 104, 106, 108, 110 and 111, each of the detector assemblies 95 being preferably located in the center of the zone 81 being defined thereby. Similarly, the six columns 102 through 111, inclusive, of detectors 95 are preferably arrayed on the top portion 88 of the helmet housing 82 so as to also provide six rows 112, 114,

116, 118, 120 and 122 of detector assemblies 95, each detector assembly 95, once again, being preferably located in the center of the detection zone 81 defined thereby. In this manner a six-by-six array of detector assemblies 95 is provided so as to provide 36 radiation emission viewing surfaces for the detection zones 81 in the plane of the top portion 88 of housing 82, each of such surfaces being the associated detection zone surface of entry of view into the brain volume for the associated detector assembly 95. Similarly, front and rear portions 84 and 86 and side portions 90 and 92 each are preferably shown as having four rows of six detector assemblies 95 each, although other arrangements may be utilized, each detector assembly 95 also preferably being located in the center of the detection zone 81 defined thereby, so as to provide a four-by-six array of 24 detector assemblies 95 in the plane of the respective helmet portion 84, 86, 90 and 92 so as to provide 24 such viewing surfaces for the detection zones 81 in each of these planes. This provides 132 of such radiation emission viewing surfaces for the preferred 144 detection zones 81. The additional twelve detector assemblies 95 which are required so as to provide the preferred equal plurality of 144 detector assemblies 95 are provided by means of modular banks 95 and 98 which each include a row of six detector assemblies 95. It should be understood that many of the detectors as enumerated may, if desired, be relocated along edges of the housing 82, as required, to yield the preferred 144 independent linear equations. The preferred orientation and alignment of the headgear apparatus 66 with respect to the brain volume is preferably identical with that previously described with reference to the preferred method of the present invention as is the method of determining the arterial blood radiation concentration.

Now referring to the operation of the system 10 in accordance with the alternative method of the present invention so as to provide the brain tissue radioisotope concentration quantity C_t for, preferably, each of the detection zones 81 so that this data may be processed in accordance with the previously stated Fick-Kety expression. The radioactive gas, which as was previously mentioned dissolves in the blood passing through the lungs, is delivered to the brain parts blood detection zones 81 in proportion to the local rate of blood flow. The resultant single energy level gamma radiation emissions detected by the various detector assemblies 95 included in the headgear apparatus 66 detector array are a function of the amount of radioactive gas deposited, or radioisotope concentration, in the various detection zones 81 and the effective field of view associated with a given detector assembly 95.

The detector assemblies 95 "look" at the brain from 144 different positions (or locations), and, as was previously mentioned, the degree of resolution or localization observed by each detector assembly 95, that is, its effective field of view, is determined by the collimator 140 associated with the particular detector 142. As was also previously mentioned, in order to properly determine the rate of local cerebral blood flow, the radioactive gas must be administered to the patient in sufficiently high concentration to provide a statistically acceptable counting rate, or quantity of detectable radiation emission, in the preferred sampling interval which is 10 seconds, but not so high that the patient is endangered by the radiation. The preferred square lumen collimator 140 utilized in accordance with the alternative

method of the present invention has a length which preferably provides a three detection zone **81** useful depth of view for the effective field of view, which field of view includes a proportionate amount of such zones **81** which are in a three-by-three matrix array in a plane normal to the direction of view for both the second and third zones in depth in a straight line, with these zones being at the center of each three-by-three matrix array. Such a collimator **140** represents the optimal compromise between an acceptable level of radioactive gas isotope administration and the collimator size suitable for localization. Each detector **142**, therefore, sees this effective field of view of three detection zones in depth, the total or resultant radiation emission observed by each detector **142**, as was previously mentioned, being due to the radioisotope concentration present in the portions of each of the detection zones **81** contained within the effective field of view of the particular detector **142**.

This relationship between the resultant radiation emission detected or observed by a given detector **142** and the radioisotope concentration present in the portions of each of the detection zones **81** contained within the effective field of view of the detector **142** may be defined by the following general expression in which, for purposes of explanation, it is assumed that all detection zones **81** are included within the bone surrounding the patient's brain, although it would be obvious to one of ordinary skill how to readily modify the following expression to take into account the surrounding bone if the detection zones were defined so as to include this bone which, in such instance, could be considered as non-radioactive source bearing material. Before defining this expression, however, a nomenclature for the detection zones **81** including the effective field of view, must first be established so that this general expression may be preferably defined for a typical detection zone **81c**. The general expression may be rewritten for each of the other one hundred forty-three detection zones **81** in the array.

For purposes of explanation, based upon the effective field of view, and the preferred planes of orientation of the headgear apparatus **66**, the given detector assembly **95c**, for which the expression is defined, shall be assumed to be located, in the most general case, adjacent the center of a three-by-three zone effective matrix array in the plane normal to the direction of view, which array is three zones in depth from that plane so as to provide a three-by-three-by-three matrix array effective volume comprising 27 detection zones **81**. Such an array is shown in FIG. 7A which is an exploded perspective view of this mathematical model matrix array, the letter "X" designating the center detection zone **81c** associated with this detector assembly **95c**. Each detection zone within this effective matrix array will be defined by three coordinates of position in the direction of view these coordinates being depth, row, and column, in that order, where the column is in a vertical direction normal to the direction of view and the row is in a horizontal direction normal to the direction of view; wherein A represents the designator for the plane of the first zone in depth; B represents the designator for the parallel plane of the second zone in depth; C represents the designator for the parallel plane of the third zone in depth; "O" represents the designator for the center row in each of the parallel planes (A, B, C) of the zones in depth; "T" represents the designator for each of the zones in the row above the center row in

each of these planes (A, B, C) in depth; "U" represents the designator for each of the zones in the row below the center row in each of these planes (A, B, C) in depth; "O" represents the designator for the center column in each of these planes (A, B, C) in depth; "R" represents the designator for each of the zones in the column to the right of the center column in each of these planes (A, B, C) in depth; and "L" represents the designator for each of the zones in the column to the left of the center column of each of these planes (A, B, C) in depth. Utilizing this labeling system or nomenclature of designators, the zones **81** in the effective matrix array are designated as shown in FIG. 7A, with "AOO" being the designation or coordinates describing the center detection zone **81c** labeled "X" for the general case.

In addition, as shown in FIG. 8, which is a diagrammatic view of the useful depth of view in the effective field of view for the typical detector assembly **95c** having a collimator **140c** and a detector **142c**, each detection zone **81** in this alternative method of the present invention is preferably further subdivided in depth in the direction of view **200** into equal volume thirds designated 1, 2 and 3 for the first zone in depth (bracket A); 4, 5 and 6 for the second zone in depth (bracket B) in the direction or axis of view **200**; and 7, 8 and 9 for the third zone in depth (bracket C) in the direction of view **200**. For purposes of explanation, it shall be assumed that the radiation concentration contained in each third of a detection zone 1 through 9, inclusive, may be represented by a point source at the center of each of the detection zone thirds 1 through 9, inclusive, along the direction or axis of view **200**, the effective field of view preferably being symmetrical about this axis **200**. These central point source locations are designated a, b, c, e, f, g, h, j, and k, for detection zone thirds 1 through 9, inclusive, respectively.

Since, as was previously mentioned, the effective field of view is preferably symmetrical about the axis of view **200**, utilizing the above defined nomenclature, we can thereby define the following equalities with respect to the geometric proportionate amount of the various detection zones contained within the effective field of view of zone "X" **81c** which zone "X" **81c** has coordinates AOO: BRO=BOT=BLO=BOU; BRU=A BLU=BRT=BLT; CRO=COT=CLO=COU; and CRU=CLU=CLT=CLT. It should be noted that these equalities relate only to the geometric proportionate amounts of these zones which are present and not to the various radiation concentrations which are present in these zones, which concentrations may differ. In addition, the collimator **140c** is preferably designed so that the effective field of view does not include ARO, ALO, ART, AOT, ALT, ARU, AOU and ALU; in other words, it includes no other zones than AOO in plane A.

Now that we have defined our nomenclature, and taking into account the above identified equalities and the previously mentioned assumptions, the general expression defining the relationship between the resultant radiation emission detected by a given detector **142c** at position AOO, which is the starting zone or point of entry of its effective field of view into the array, and the radiation concentration present in the portions of each of the detection zones **81** contained within the effective field of view of detector **142c** in accordance with the alternative method of the present invention is as follows, letting " $P_{T,AOO}$ " represent the detected resultant radiation emission or counting rate resulting from the

contributions of all the detection zones 81 within the effective field of view; " ϵ " represents the efficiency of the detection system; " ϵ " represent the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination (loss due to radioactive decay); " s " represent the factor expressing the attenuation of radiation by the scalp and skull; " A_d " represent the area of the detector 142c; " K " represent the dimensional constant; " μ " represent the total attenuation constant for gamma radiation of a specific energy (in square centimeters/grams); " t " represents the thickness of brain tissue through which the radiation is attenuated (in grams/square centimeter) in multiples of one-sixth of a zone; " l " represent the distance from the center of

each one-third of a zone to the detector 142c along the axis of view 200; " V_T " represent the volume of each one-third of a zone; " V POSITION" represent the effective pyramidal intercept of volume for one-third of a zone whose positional coordinates are represented by the three proper coordinates for a given "POSITION" (see FIG. 7). " V POSITION", including compensation for the collimator factor, which is the projected area of the detector 142c divided by the total area of the detector 142c as seen through the collimator 140c; and " I POSITION" represent the radiation concentration (in microcuries) of radioisotope contained within the one-third of a zone whose positional coordinates are represented by the three proper coordinates for a given "POSITION" (see FIG. 7):

$$\begin{aligned}
 P_{T,100} = \frac{KA_d \epsilon \delta s}{V_T} & \left[I_{A,00} V_{A,00} \frac{e^{-\mu t_{0-a}}}{l_a^2} + I_{A_2,00} V_{A_2,00} \frac{e^{-\mu t_{0-b}}}{l_b^2} + \right. \\
 & I_{A_3,00} V_{A_3,00} \frac{e^{-\mu t_{0-c}}}{l_c^2} + I_{B_4,00} V_{B_4,00} \frac{e^{-\mu t_{0-d}}}{l_d^2} + I_{B_5,00} V_{B_5,00} \frac{e^{-\mu t_{0-e}}}{l_e^2} + \\
 & I_{B_6,00} V_{B_6,00} \frac{e^{-\mu t_{0-f}}}{l_f^2} + I_{C_7,00} V_{C_7,00} \frac{e^{-\mu t_{0-g}}}{l_g^2} + I_{C_8,00} V_{C_8,00} \frac{e^{-\mu t_{0-h}}}{l_h^2} + I_{C_9,00} V_{C_9,00} \frac{e^{-\mu t_{0-i}}}{l_i^2} + \\
 & I_{C_9,00} V_{C_9,00} \frac{e^{-\mu t_{0-k}}}{l_k^2} + V_{B_4,0T} \frac{e^{-\mu t_{0-e}}}{l_e^2} (I_{B_4,0T} + I_{B_4,0R} + I_{B_4,0L} + I_{B_4,0U}) \\
 & + V_{B_5,0T} \frac{e^{-\mu t_{0-f}}}{l_f^2} (I_{B_5,0T} + I_{B_5,0R} + I_{B_5,0L} + I_{B_5,0U}) + \\
 & V_{B_6,0T} \frac{e^{-\mu t_{0-g}}}{l_g^2} (I_{B_6,0T} + I_{B_6,0R} + I_{B_6,0L} + I_{B_6,0U}) + V_{C_7,0T} \frac{e^{-\mu t_{0-h}}}{l_h^2} (I_{C_7,0T} \\
 & + I_{C_7,0R} + I_{C_7,0L} + I_{C_7,0U}) + V_{C_8,0T} \frac{e^{-\mu t_{0-i}}}{l_i^2} (I_{C_8,0T} + I_{C_8,0R} + \\
 & I_{C_8,0L} + I_{C_8,0U}) + V_{C_9,0T} \frac{e^{-\mu t_{0-k}}}{l_k^2} (I_{C_9,0T} + I_{C_9,0R} + I_{C_9,0L} + \\
 & I_{C_9,0U}) + V_{B_4,1T} \frac{e^{-\mu t_{0-e}}}{l_e^2} (I_{B_4,1T} + I_{B_4,1R} + I_{B_4,1L} + I_{B_4,1U}) + \\
 & V_{B_5,1T} \frac{e^{-\mu t_{0-f}}}{l_f^2} (I_{B_5,1T} + I_{B_5,1R} + I_{B_5,1L} + I_{B_5,1U}) + \\
 & V_{B_6,1T} \frac{e^{-\mu t_{0-g}}}{l_g^2} (I_{B_6,1T} + I_{B_6,1R} + I_{B_6,1L} + I_{B_6,1U}) + \\
 & V_{C_7,1T} \frac{e^{-\mu t_{0-h}}}{l_h^2} (I_{C_7,1T} + I_{C_7,1R} + I_{C_7,1L} + I_{C_7,1U}) + \\
 & V_{C_8,1T} \frac{e^{-\mu t_{0-i}}}{l_i^2} (I_{C_8,1T} + I_{C_8,1R} + I_{C_8,1L} + I_{C_8,1U}) + \\
 & \left. V_{C_9,1T} \frac{e^{-\mu t_{0-k}}}{l_k^2} (I_{C_9,1T} + I_{C_9,1R} + I_{C_9,1L} + I_{C_9,1U}) \right]
 \end{aligned}$$

When all the constants for the system 10 at the time of the data collection are solved for and appropriate terms combined, and letting C_1 through C_7 represent these constants, this expression may be reduced to:

$$P_{T, AOO} = C_1 I_{AOO} + C_2 I_{BOO} + C_3 I_{COO} + C_4 (I_{ROO} + I_{BRO} + I_{BLO} + I_{ROU}) + C_5 (I_{COT} + I_{CRO} + I_{CLO} + I_{COT}) + C_6 (I_{BRT} + I_{BRT} + I_{BRT} + I_{BLU}) + C_7 (I_{CRT} + I_{CLT} + I_{CRU} + I_{CLU})$$

The above expanded general expression for P_T , as can be observed, includes corrections for tissue attenuation of the gamma radiation originating at various depths within the brain; i.e., in other detection zones, as well as for the inverse square relationship between counts detected and distance.

It should be noted that taking the above general expression, in conjunction with FIGS. 7A, 7B and 8 and providing each actual detection zone with a unique identifier, such as Z-1 through Z-144, inclusive (FIG. 7B) this expression can then be readily rewritten by inspection by anyone with ordinary mathematical skill for any one of the 144 detection zones 81 in the array by assuming it to be located at this center position relative to its effective field of view and substituting the equivalent identifiers Z-1 through Z-144, inclusive, which correspond to the above subscripts which relate to the position of this zone and the portions of the zones contained within its effective field of view, the effective fields of the view being identical for each detector 142. For example, if the detection zones 81 are uniquely identified by numbering consecutively left-to-right, row-by-row starting at the top row as shown in FIG. 7B, and a given detector is located by position adjacent the zone designated Z-8, where AOO=Z-8, ALO=Z-7, ARO=Z-9, ALU=Z-13, AOU=Z-14, ARU=Z-15, ALT=Z-1, AOT=Z-2, ART=Z-3, BOO=Z-32, BLO=Z-31, BRO=Z-33, BLUZ-37, BOW=Z-38, BRU=Z-39, BLT=Z-25, BOF=Z-26, BRT=Z-27, COO=Z-56, CLO=Z-55, CRO=Z-57, CLU=Z-61, COU=Z-62, CRU=Z-63, CLT=Z-49, COT=Z-50, and CRT=Z-51, the above described general expression for P_T , the detected resultant radiation emission, assuming the constants for the system 10 at the time of the data collection to have been solved for and appropriate terms combined, for the detector 142 adjacent zone Z-8, would be rewritten as:

$$P_{T, Z-8} = C_1 I_{Z-8} + C_2 I_{Z-32} + C_3 I_{Z-56} + C_4 (I_{Z-26} + I_{Z-33} + I_{Z-31} + I_{Z-38}) + C_5 (I_{Z-50} + I_{Z-57} + I_{Z-55} + I_{Z-62}) + C_6 (I_{Z-27} + I_{Z-25} + I_{Z-39} + I_{Z-37}) + C_7 (I_{Z-51} + I_{Z-49} + I_{Z-63} + I_{Z-61})$$

It should further be noted that some of the terms in the general expression may be zero when its expression is written for a detector 142 whose effective field of view contains a portion or portions which fall outside the 144 detection zone array into which the brain is preferably subdivided, such as detector which is located at a corner or along the outer periphery of the array (such as at position Z-1 in FIG. 7B in the example given). In such instance the effective 27 zone matrix array, only 19 of which are "seen", will have portions of zones that do not exist as actual predefined detection zones within the effective field of view within the subdivided brain volume. For example, utilizing the same numbering system (FIG. 7B) as in the previous example, if the given detector is located by position adjacent the zone designated Z-1, which is a corner, where the following equivalents now exist, the numeral "0" indicating that the zone does not exist, AOO=Z-1, ALO=0, ARO=Z-2, ALU=0, AOU=Z-7, ARU=Z-8, ALT=0, AOT=0, AOT=0, ART=0, BOO=Z-25, BLO=0, BRO=Z-26, BLU=0, BOU=Z-31, BRU=Z-

32, BLT=0, BOT=0, BRT=0, COO=Z-49, CLO=0, CRO=Z-50, CLU=0, COU=Z-55, CRU=Z-56, CLT=0, COT=0, and CRT=0, the above described general expression for P_T for the detector 142 adjacent zone Z-1 would be rewritten as:

$$P_{T, Z-1} = C_1 I_{Z-1} + C_2 I_{Z-25} + C_3 I_{Z-49} + C_4 (0 + I_{Z-26} + 0 + I_{Z-31}) + C_5 (0 + I_{Z-50} + 0 + I_{Z-55}) + C_6 (0 + 0 + I_{Z-32} + 0) + C_7 (0 + 0 + I_{Z-56} + 0)$$

which reduces to

$$P_{T, Z-1} = C_1 I_{Z-1} + C_2 I_{Z-25} + C_3 I_{Z-49} + C_4 (I_{Z-26} + I_{Z-31}) + C_5 (I_{Z-50} + I_{Z-55}) + C_6 I_{Z-32} + C_7 I_{Z-56}$$

Similarly, after a system of providing each detection zone 81 with a unique identifier Z-1 through Z-144, inclusive, is established, the expression for P_T can therefore, as was previously mentioned, be rewritten by inspection for each detector 142 in the array thereby preferably providing 144 counting expressions P_T for the various parts of the same tissue volume. Since, as can be observed from the foregoing general expression for P_T , the quantity of detected radiation emission is directly related to the tissue radioisotope concentrations in the portions of the detection zones contained within the effective field of view, the tissue radioisotope concentration quantity "C", which is equivalent to "I" for a given detection zone, for each of the individual detection zones can be determined from the actual resultant radiation emissions observed by the detector array of the helmet housing 82, by solving these 144 equations or expressions for P_T simultaneously.

The processing of this data as well as the data collection thereof is preferably identical with that previously described with reference to the preferred method of the present invention with the exception that the computer 61 is preferably programmed in conventional fashion to solve the 144 simultaneous equations for each 10-second period rather than four sets of 36 simultaneous equations each as when the brain volume is divided into quadrants. Thus, the summary of the alternative method of the present invention is preferably identical with that previously described with reference to the preferred method of the present invention except for the counting expression utilized and the above difference in processing of the data.

By utilizing the method and apparatus of the present invention local cerebral blood flow can be determined in a rapid and efficient manner with minimal discomfort to the patient. Furthermore, this information can be provided in any desired format which will enhance its usefulness as a valuable diagnostic tool for the physician.

It is to be understood that the above described embodiments of the invention are merely illustrative of the principles thereof and that numerous modifications and embodiments of the invention may be derived within the spirit and scope thereof, such as by dissolving the concentration of a radioactive substance in the blood by other than inhalation of a radioactive gas or by utilizing arterial catheterization, such as of the carotid artery, to determine the arterial blood radioisotope concentration, although these alternatives are not as desirable as the preferred or described alternative method. In addition, it should be understood that an exact integration may be substituted for the approximations utilized in the counting expression without departing from the present invention.

What is claimed is:

1. A method for determining a discrete localized blood flow in a preselected radiation emission detection zone of a volume of tissue comprising the steps of

a. defining a plurality of radiation emission detection zones for at least a portion of said tissue volume, said plurality of zones including said preselected zone;

b. defining a plurality of effective fields of view, each of said plurality of zones being within at least one effective field of view, each of said effective fields of view being defined with respect to an associated predetermined axis of view, each of said effective fields of view including said preselected zone and at least a portion of one or more of said other zones of said plurality of zones;

c. providing a source of radiation emission for said blood for causing each detection zone to have an associated quantity of radioisotope concentration and an associated quantity of radiation emission dependent on said associated radioisotope concentration; and

d. quantifying said associated quantity of radioisotope concentration for said preselected zone by quantifying a resultant radiation emission quantity for each of said effective fields of view, said resultant radiation emission quantities being dependent on said radioisotope concentrations of said preselected zone and said portions of said plurality of zones included within each of said effective fields of view, whereby said discrete localized blood flow may be determined.

2. A method in accordance with claim 1 wherein said detection zone defining step comprises defining a plurality of radiation emission detection zones for substantially all of said tissue volume, said plurality of zones including said preselected zone.

3. A method in accordance with claim 1 wherein said detection zone defining step further comprises defining said detection zones in at least a quadrant of said tissue volume, said radiation emission providing step comprises providing a source of radiation emission having at least two distinct radiation emission energy levels, said effective field defining step comprises defining a different effective field of view for each of said energy levels, and said quantifying step comprises quantifying said associated quantity of radioisotope concentration for said preselected zone by quantifying a resultant radiation emission quantity for each of said effective fields of view.

4. A method for determining a local cerebral blood flow in a preselected radiation emission detection zone of a brain comprising the steps of

a. defining a plurality of cerebral radiation emission detection zones for at least a portion of said brain, said plurality of zones including said preselected zone;

b. defining a plurality of effective fields of view, each of said plurality of zones being within at least one effective field of view, each of said effective fields of view being defined with respect to an associated predetermined axis of view, each of said effective fields of view including said preselected zone and at least a portion of one or more of said other zones of said plurality of zones;

c. providing a source of radiation emission for said cerebral blood for causing each detection zone to have an associated quantity of radioisotope concentration and an associated quantity of radiation emission dependent on said associated radioisotope concentration; and

d. quantifying said associated quantity of radioi-

sotope concentration for said preselected zone by quantifying a resultant radiation emission quantity for each of said effective fields of view, said resultant radiation emission quantities being dependent on said radioisotope concentrations of said preselected zone and said portions of said plurality of zones included within each of said effective fields of view, whereby said local cerebral blood flow may be determined.

5. A method in accordance with claim 4 wherein said detection zone defining step comprises defining a plurality of cerebral radiation emission detection zones for substantially all of said brain, said plurality of zones including said preselected zone.

6. A method in accordance with claim 4 wherein the step of providing a source of radiation emission for said cerebral blood includes the step of dissolving a concentration of a radioactive substance in said cerebral blood over a predetermined period of time, said radioactive substance having an associated radiation emission, said zone associated quantities of radiation emission and radioisotope concentration being dependent upon said substance associated radiation emission and radioisotope concentration.

7. A method in accordance with claim 6 wherein the dissolving step includes the additional step of dissolving an increasing concentration of said radioactive substance in said cerebral blood over said predetermined period of time.

8. A method in accordance with claim 7 wherein the step of quantifying said associated quantities of radiation emission includes the step of quantifying said associated quantities of radiation emission during said predetermined time.

9. A method in accordance with claim 6 wherein said dissolving step includes the additional step of inhaling a radioactive gas in a closed inhalation-exhalation breathing system over said predetermined period of time.

10. A method in accordance with claim 9 wherein said additional inhalation step includes the further step of injecting a predetermined quantity of said gas into said closed system substantially at the beginning of an inhalation by said patient.

11. A method in accordance with claim 10 wherein said further injecting step includes a still further step of injecting a predetermined quantity of said gas into said closed system substantially at the beginning of each inhalation by said patient during said period of time whereby said cerebral blood gas concentration is increased over said predetermined period of time.

12. A method in accordance with claim 10 wherein said further injecting step includes the still further step of mixing said injected gas with an immediately preceding exhalation of said patient prior to said inhalation by said patient.

13. A method in accordance with claim 9 wherein said additional inhaling step includes the further step of inhaling a radioactive gas which is substantially inert to said cerebral blood.

14. A method in accordance with claim 9 wherein said inhaling step includes the further step of inhaling a radioactive gas having an associated gamma radiation emission, said gamma emission being said quantified radiation emission.

15. A method in accordance with claim 9 wherein said additional inhaling step includes the further step of inhaling Xenon¹³³ as said radioactive gas.

16. A method in accordance with claim 9 wherein said additional inhaling step includes the further step of inhaling Krypton⁸⁵ as said radioactive gas.

17. A method in accordance with claim 4 wherein said zone defining step includes the additional step of defining at least 144 radiation emission detection zones in said plurality of zones for defining substantially all of said brain.

18. A method in accordance with claim 4 wherein said zone defining step includes the additional step of defining a multiplanar array of said detection zones in said plurality of zones.

19. A method in accordance with claim 4 wherein said zone defining step includes the further step of defining a reproducible basal plane for said brain as a reference plane, said plurality of detection zones being defined with respect to said basal plane.

20. A method in accordance with claim 19 wherein said further zone defining step includes the still further step of defining the glabellar-sellar plane of the brain as said basal plane.

21. A method in accordance with claim 20 wherein said still further zone defining step includes the additional step of defining the mid-sagittal plane of the brain as a second reference plane for said brain, said second reference plane being substantially perpendicular to said basal plane, said plurality of detection zones being further defined with respect to said second reference plane.

22. A method in accordance with claim 21 wherein said still further zone defining step includes the still ad-

ditional step of defining a third reference plane for said brain substantially perpendicular to said basal plane and said second reference plane, and tangent to the forehead, said plurality of detection zones being still further defined with respect to said third reference plane.

23. A method in accordance with claim 21 wherein said reference plane is defined as a zero reference plane, said detection zones being defined with respect to said zero reference plane.

24. A method in accordance with claim 23 wherein said zone defining step includes the step of defining said detection zones by a multiplanar array of radiation emission detection means arrayed about said brain with respect to said zero reference plane.

25. A method in accordance with claim 24 wherein each of said detection zones is defined as being of substantially equal dimension.

26. A method in accordance with claim 25 wherein said multiplanar array is a substantially rectangular multiplanar array defining at least 144 detection zones for defining substantially all of said brain.

27. A method in accordance with claim 4 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified in accordance with the expression

$$\begin{aligned}
 P_{T,ACO} = \frac{KA_d \epsilon \delta s}{V_T} & \left[I_{A,00} V_{A,00} \frac{e^{-\mu t_{0a}}}{l_a^2} + I_{A,00} V_{A,00} \frac{e^{-\mu t_{0b}}}{l_b^2} + \right. \\
 & I_{A,00} V_{A,00} \frac{e^{-\mu t_{0c}}}{l_c^2} + I_{B,00} V_{B,00} \frac{e^{-\mu t_{0e}}}{l_e^2} + I_{B,00} V_{B,00} \frac{e^{-\mu t_{0f}}}{l_f^2} + \\
 & I_{B,00} V_{B,00} \frac{e^{-\mu t_{0g}}}{l_g^2} + I_{C,00} V_{C,00} \frac{e^{-\mu t_{0h}}}{l_h^2} + I_{C,00} V_{C,00} \frac{e^{-\mu t_{0j}}}{l_j^2} + \\
 & I_{C,00} V_{C,00} \frac{e^{-\mu t_{0k}}}{l_k^2} + V_{B,OT} \frac{e^{-\mu t_{0e}}}{l_e^2} (I_{B,OT} + I_{B,RO} + I_{B,LO} + I_{B,OU}) \\
 & + V_{B,OT} \frac{e^{-\mu t_{0f}}}{l_f^2} (I_{B,OT} + I_{B,RO} + I_{B,LO} + I_{B,OU}) + \\
 & V_{B,OT} \frac{e^{-\mu t_{0g}}}{l_g^2} (I_{B,OT} + I_{B,RO} + I_{B,LO} + I_{B,OU}) + V_{C,OT} \frac{e^{-\mu t_{0h}}}{l_h^2} (I_{C,OT} \\
 & + I_{C,RO} + I_{C,LO} + I_{C,OU}) + V_{C,OT} \frac{e^{-\mu t_{0j}}}{l_j^2} (I_{C,OT} + I_{C,RO} + \\
 & I_{C,LO} + I_{C,OU}) + V_{C,OT} \frac{e^{-\mu t_{0k}}}{l_k^2} (I_{C,OT} + I_{C,RO} + I_{C,LO} + \\
 & I_{C,OU}) + V_{B,LT} \frac{e^{-\mu t_{0e}}}{l_e^2} (I_{B,LT} + I_{B,RO} + I_{B,LO} + I_{B,RT}) + \\
 & V_{B,LT} \frac{e^{-\mu t_{0f}}}{l_f^2} (I_{B,LT} + I_{B,RO} + I_{B,LO} + I_{B,RT}) +
 \end{aligned}$$

$$\begin{aligned}
& \frac{V_{B_6LT} e^{-\mu t_{0-g}}}{l_g^2} (I_{B_6LT} + I_{B_6RU} + I_{B_6LU} + I_{B_6RT}) + \\
& \frac{V_{C_7LT} e^{-\mu t_{0-h}}}{l_h^2} (I_{C_7LT} + I_{C_7RU} + I_{C_7LU} + I_{C_7RT}) + \\
& \frac{V_{E_8LT} e^{-\mu t_{0-j}}}{l_j^2} (I_{E_8LT} + I_{E_8RU} + I_{E_8LU} + I_{E_8RT}) + \\
& \frac{V_{G_9LT} e^{-\mu t_{0-k}}}{l_k^2} (I_{G_9LT} + I_{G_9RU} + I_{G_9LU} + I_{G_9RT}) \Big]
\end{aligned}$$

for a detection zone whose position is designated A00 20
for the positional coordinate system shown in FIG. 7A,
where the subscript 1, 2 and 3, respectively, designate
consecutive thirds of a detection zone in the A plane of
said system; 4, 5 and 6, respectively, designate consec- 25
utive thirds of a detection zone in the B plane of said
system; 7, 8 and 9, respectively, designate consecutive
thirds of a detection zone in the C plane of said system;
a, b, c, e, f, g, h, j and k represent the effective central
point source locations of radiation concentration for
the detection zone thirds 1, 2, 3, 4, 5, 6, 7, 8 and 9, res- 30
pectively; the associated effective field of view is sym-
metrical with respect to said associated axis so as to
provide an effective geometric relationship for said sys-
tem of BRO=BOT=BLO=BOU, BRU= λ
BLU=BRT=BLT, CRO=COT=CLO=COU, and 35
CRU=CLU=CRT=CLT; $P_{T,400}$ represents the
associated quantified resultant radiation emission
quantity; ϵ represents the efficiency of said detec-
tion means; δ represents the fraction of source re- 40
maining since its original assay at the time of quanti-
fication; s represents the factor expressing the
attenuation of radiation by the patient's scalp and
skull; A_d represents the area of said associated
detector; K represents a dimensional constant; 45
 μ represents the total attenuation constant for radia-
tion emission of a specific energy; t represents the
thickness of brain tissue through which the radiation
emission is attenuated in multiples of one-sixth of a
detection zone; l represents the distance from the center 50
of each one-third of a detection zone to the associated
detector along the associated axis of view; V_T repre-
sents the volume of each one-third of a detection zone;
 V_{POSITION} , where the proper associated positional
coordinates are substituted for the subscript term PO- 55
SITION in the expression, represents the effective in-
tercept of volume for the associated third of a detec-
tion zone, and I_{POSITION} , where the proper associated
positional coordinates are substituted for the subscript
term POSITION in the expression, represents the radia- 60
tion concentration for the associated one-third of a de-
tection zone.

28. A method in accordance with claim 27 wherein
said still further quantifying step includes the additional
step of quantifying each of said resultant quantities of
radiation emission in accordance with said expression 65
so as to provide a plurality of simultaneous equations
in which at least one equation is provided for each re-
spective zone included within said plurality of effective

fields of view.

29. A method in accordance with claim 28 wherein
said quantifying step includes the still further step of
processing said simultaneous equations in a data pro-
cessing means to provide a tissue radiation concentra-
tion characteristic for said preselected detection zone.

30. A method in accordance with claim 4 wherein
said method includes the additional step of performing
step (d) to provide a first tissue radioisotope concentra-
tion as said associated radioisotope concentration
for said preselected detection zone and, (e) thereafter
repeating step (d) to provide a second tissue radioi-
sotope concentration as said associated radioisotope
concentration for said preselected detection zone so as
to provide a tissue radioisotope concentration charac-
teristic for said preselected detection zone which is an
indication of a rate of change in said tissue radioisotope
concentration for said preselected detection zone, said
local cerebral blood flow determination being depend-
ent thereon.

31. A method in accordance with claim 6 wherein
said dissolving step includes the additional step of in-
haling a radioactive gas over said predetermined time;
and said method includes the additional step of (f)
quantifying a radioisotope concentration of said radio-
active gas in arterial cerebral blood during said prede-
termined time, said local cerebral blood flow determi-
nation being dependent on said arterial blood gas radi-
oisotope concentration.

32. A method in accordance with claim 31 wherein
said gas radioisotope concentration quantifying step
includes the further step of quantifying said gas radioi-
sotope concentration in the last part of an exhalation
of said patient during said predetermined time, so as to
provide an arterial blood radioisotope concentration
characteristic.

33. A method in accordance with claim 32 wherein
said further gas radioisotope concentration quantifying
step includes the still further step of quantifying said
gas radioisotope concentration in the last part of each
exhalation of said patient during said predetermined
time so as to provide an arterial blood radioisotope
concentration characteristic which is an indication of
a rate of change in said arterial blood concentration,
said local cerebral blood flow determination being de-
pendent on said arterial blood concentration character-
istic.

34. A method in accordance with claim 32 wherein
said further gas radioisotope concentration quantifying

step includes the still further step of end-tidal sampling said exhalation in cooperative relation with said quantifying of said associated quantity of radioisotope concentration for said preselected detection zone.

35. A method in accordance with claim 6 wherein said dissolving step includes the additional step of inhaling a radioactive gas over said predetermined time; and said method includes the further step of quantifying the gas concentration of carbon dioxide in the last part of an exhalation of said patient during at least a portion of said predetermined time, said local cerebral blood flow determination being dependent on said carbon dioxide concentration.

36. A method in accordance with claim 35 wherein said further gas concentration quantifying step includes the still further step of quantifying the gas concentration of oxygen in said exhalation last part, said local cerebral blood flow determination being further dependent on said oxygen concentration.

37. A method in accordance with claim 6 wherein said dissolving step includes the additional step of inhaling a radioactive gas over said predetermined time; and said method includes the further step of quantifying the gas concentration of oxygen in the last part of an exhalation of said patient during at least a portion of said predetermined time, said local cerebral blood flow determination being dependent on said oxygen concentration.

38. A method in accordance with claim 4 wherein said associated radioisotope concentration is a tissue radioisotope concentration and said method includes the additional step of repeating step (d) at 10 second intervals over a period of a minute so as to provide a tissue radioisotope concentration characteristic which is an indication of rate of change in said tissue radioisotope concentration, said local cerebral blood flow determination being dependent on said tissue radioisotope concentration characteristic.

39. A method in accordance with claim 38 wherein said providing step includes the further step of providing said source of radiation emission by dissolving a concentration of a radioactive gas in said blood by inhalation of said gas in a closed inhalation-exhalation system over said one minute period, and said method includes the additional step of quantifying a radioisotope concentration of said gas in the last part of an exhalation of said patient at said ten second intervals so as to provide an arterial blood radioisotope concentration characteristic which is an indication of rate of change in said arterial blood concentration, said local cerebral blood flow determination being further dependent on said arterial blood concentration characteristic.

40. A method in accordance with claim 4 wherein said quantifying step includes the step of providing a tissue radioisotope concentration quantity as said associated radioisotope concentration for said preselected detection zone in accordance with said quantified resultant radiation emission quantities; said radiation emission source providing step includes the step of dissolving a radioactive substance in said blood by inhalation of a radioactive gas; and said method includes the additional steps of providing an arterial blood radioisotope concentration quantity for said dissolved gas, and determining said local cerebral blood flow in accordance with the expression

$$\int_{m_1}^{m_2} \dot{C}_i W dt = F \int_{m_1}^{m_2} (\dot{C}_a - \dot{C}_v) dt$$

where C_i equals the brain tissue radioisotope concentration quantity in $\mu C/g$, $(C_a - C_v)$ equals the arterial-venous blood radioisotope concentration difference in $\mu C/g$, m_1 equals the starting time of said determination in seconds, F equals the local cerebral blood flow in $\mu l/g/min$, W equals the weight of the tissue comprising the detection zone, and m_2 equals the number of seconds after said inhalation is commenced.

41. A method in accordance with claim 6 wherein said dissolving step includes the additional step of inhaling a radioactive gas over said predetermined period of time.

42. A method in accordance with claim 4 wherein said detection zone defining step comprises defining a plurality of radiation emission detection zones for substantially all of said brain volume, said plurality of zones including said preselected zone.

43. A method in accordance with claim 42 wherein said detection zone defining step further comprises defining said detection zones in a plurality of quadrants of said brain volume and said quantifying step further comprises quantifying said associated quantity of radioisotope concentration for said quadrants by independently quantifying a resultant radiation emission quantity for each of said effective fields of view in said one quadrant apart from said other quadrants.

44. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$Q_{G4} = \bar{m} I_4 + \bar{n} I_{13}$$

$$Q_{G1} = \bar{m} I_1 + \bar{n} I_4$$

$$Q_{F1} = \bar{p} I_1 + \bar{q} I_4 + \bar{g} I_{10} + \bar{r} I_{13}$$

$$Q_{C10} = \bar{m} I_{10} + \bar{n} I_{13}$$

for a typical quadrant as illustrated in FIGS. 9A-9E, where $Q_{\text{subscript}}$ represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 9A-9E, I_{zone} represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{m} = C \int_0^{V_{1Q\text{ZONE}}} \frac{e^{-\mu t_{1Q\text{ZONE}}}}{l_{1Q\text{ZONE}}^2} f_{C1Q\text{ZONE}} dV,$$

$$\bar{n} = C \int_0^{V_{2Q\text{ZONE}}} \frac{e^{-\mu t_{2Q\text{ZONE}}}}{l_{2Q\text{ZONE}}^2} f_{C2Q\text{ZONE}} dV,$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^z} f_{c1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^z} f_{c2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^z} f_{c2''QZONE} dV, \quad 15$$

20

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone. 35

45. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$Q_{G2} = \bar{m}_2 + \bar{n}_{11}$$

$$Q_{G3} = \bar{m}_3 + \bar{n}_{12}$$

$$Q_{A12} = \bar{m}_{12} + \bar{n}_{11}$$

$$Q_{H3} = \bar{p}_3 + \bar{q}_2 + \bar{q}_{12} + \bar{r}_{11}$$

for a typical quadrant as illustrated in FIGS. 10A-10E, where Q subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 10A-10E, I zone represents the radiation concentration of radioisotope contained within a given zone, 55

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^z} f_{c1QZONE} dV, \quad 60$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^z} f_{c2QZONE} dV, \quad 65$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^z} f_{c1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^z} f_{c2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^z} f_{c2''QZONE} dV, \quad 15$$

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where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone. 35

46. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$Q_{G8} = \bar{m}_8 + \bar{n}_{17}$$

$$Q_{G9} = \bar{m}_9 + \bar{n}_{18}$$

$$Q_{A18} = \bar{m}_{18} + \bar{n}_{17}$$

$$Q_{H9} = \bar{p}_9 + \bar{q}_8 + \bar{q}_{18} + \bar{r}_{17}$$

for a typical quadrant as illustrated in FIGS. 11A-11E, where Q subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 11A-11E, I zone represents the radiation concentration of radioisotope contained within a given zone, 55

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^z} f_{c1QZONE} dV, \quad 60$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^z} f_{c2QZONE} dV, \quad 65$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^2} f_{C1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^2} f_{C2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^2} f_{C2''QZONE} dV, \quad 15$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

47. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$\begin{aligned} Q_{G5} &= \bar{m}_5 + \bar{n}_{14} \\ Q_{G6} &= \bar{m}_6 + \bar{n}_{15} \\ Q_{A15} &= \bar{m}_{15} + \bar{n}_{14} \\ Q_{H6} &= \bar{p}_{16} + \bar{q}_{15} + \bar{q}_{15} + \bar{r}_{14} \end{aligned}$$

for a typical quadrant as illustrated in FIGS. 12A-12D, where Q subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 12A-12D, I zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^2} f_{C1QZONE} dV, \quad 60$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^2} f_{C2QZONE} dV, \quad 65$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^2} f_{C1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^2} f_{C2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^2} f_{C2''QZONE} dV, \quad 15$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

48. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$\begin{aligned} Q_{A24} &= \bar{m}_{24} + \bar{n}_{23} \\ Q_{A21} &= \bar{m}_{21} + \bar{n}_{20} \\ Q_{B21} &= \bar{p}_{21} + \bar{q}_{20} + \bar{q}_{24} + \bar{r}_{23} \\ Q_{C20} &= \bar{m}_{20} + \bar{n}_{23} \end{aligned}$$

for a typical quadrant as illustrated in FIGS. 13A-13E, where Q subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 13A-13E, I zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^2} f_{C1QZONE} dV, \quad 60$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^2} f_{C2QZONE} dV, \quad 65$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^z} f_{c1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^z} f_{c2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^z} f_{c2''QZONE} dV, \quad 15$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

49. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$\begin{aligned} Q_{A33} &= \bar{m}l_{33} + \bar{n}l_{32} \\ Q_{A30} &= \bar{m}l_{30} + \bar{n}l_{29} \\ Q_{B30} &= \bar{p}l_{30} + \bar{q}l_{29} + \bar{q}l_{33} + \bar{r}l_{32} \\ Q_{C29} &= \bar{m}l_{29} + \bar{n}l_{32} \end{aligned}$$

for a typical quadrant as illustrated in FIGS. 14A-14D, where Q subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated Q detector illustrated in FIGS. 14A-14D, l zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^z} f_{c1QZONE} dV, \quad 50$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^z} f_{c2QZONE} dV, \quad 55$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^z} f_{c1'QZONE} dV, \quad 5$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^z} f_{c2'QZONE} dV, \text{ and } 10$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^z} f_{c2''QZONE} dV, \quad 15$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

50. In a method for determining cerebral blood flow in a patient from the dissolving of a concentration of a radioactive gas in said blood by the inhalation of said gas in a closed inhalation-exhalation breathing system over a predetermined period of time, the improvement comprising the step of injecting a predetermined quantity of said gas into said closed system substantially at the beginning of an inhalation of a tidal volume by said patient so as to increase said blood gas concentration during said predetermined period of time.

51. A method in accordance with claim 50 wherein said injection step includes the step of injecting a predetermined quantity of said gas into said closed system substantially at the beginning of each inhalation of a tidal volume by said patient during said predetermined period of time whereby said cerebral blood gas concentration is increased over said predetermined period of time.

52. A method in accordance with claim 51 wherein said injecting step includes the additional step of mixing said injected gas with an immediately preceding exhalation of a tidal volume by said patient prior to said inhalation by said patient.

53. In a method for determining cerebral blood flow in a patient from the dissolving of a concentration of a radioactive gas in said blood by the inhalation of said gas over a predetermined period of time so as to provide a good gas radioisotope concentration, the improvement comprising the step of quantifying said blood gas radioisotope concentration in the end tidal volume part of an exhalation of said patient during said predetermined time so as to provide an arterial blood concentration characteristic, the arterial blood being in

concentration equilibrium with said radio active gas contained in the patient lung alveolar during said end tidal volume exhalation part.

54. A method in accordance with claim 53 wherein said gas radioisotope concentration quantifying step includes the further step of quantifying said gas radioisotope concentration in the end tidal volume part of each exhalation of said patient during said predetermined time so as to provide an arterial blood radioisotope concentration characteristic which is an indication of a rate of change in said arterial blood concentration, said local cerebral blood flow determination being dependent on said arterial blood concentration characteristic.

55. A method in accordance with claim 53 wherein said gas radiation quantifying step includes the further step of end-tidal sampling said exhalation so as to provide said exhalation part to be quantified.

56. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$P_{49} = \bar{a}I_9 + \bar{b}I_{18} + \bar{c}I_7$$

$$P_{427} = \bar{a}I_{27} + \bar{b}I_{26} + \bar{c}I_{25}$$

for a typical quadrant as illustrated in FIGS. 15A-15C and 16A and 16B, where P subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated P detector illustrated in FIGS. 15A-15C and 16A and 16B, I zone represents the radiation concentration of radioisotope contained with a given zone,

$$\bar{a} = C \int_0^{V_{1PZONE}} \frac{e^{-\mu t_{1PZONE}}}{l_{1PZONE}^2} f_{C1PZONE} dV,$$

$$\bar{b} = C \int_0^{V_{2PZONE}} \frac{e^{-\mu t_{2PZONE}}}{l_{2PZONE}^2} f_{C2PZONE} dV, \text{ and}$$

$$\bar{c} = C \int_0^{V_{3PZONE}} \frac{e^{-\mu t_{3PZONE}}}{l_{3PZONE}^2} f_{C3PZONE} dV,$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a

specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

57. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$P_{C21} = \bar{a}I_{21} + \bar{b}I_{24} + \bar{c}I_{27}$$

for a typical quadrant as illustrated in FIGS. 17A-17C, where P subscript represents the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated P detector illustrated in FIGS. 17A-17C, I zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{a} = C \int_0^{V_{1PZONE}} \frac{e^{-\mu t_{1PZONE}}}{l_{1PZONE}^2} f_{C1PZONE} dV,$$

$$\bar{b} = C \int_0^{V_{2PZONE}} \frac{e^{-\mu t_{2PZONE}}}{l_{2PZONE}^2} f_{C2PZONE} dV, \text{ and}$$

$$\bar{c} = C \int_0^{V_{3PZONE}} \frac{e^{-\mu t_{3PZONE}}}{l_{3PZONE}^2} f_{C3PZONE} dV,$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

58. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$P_{H12} = \bar{d}l_{12} + \bar{e}l_{11} + \bar{e}l_{15} + \bar{n}l_{14} + \bar{g}l_{13} + \bar{g}l_{17} + \bar{h}l_{16}$$

$$Q_{C19} = \bar{m}l_{19} + \bar{n}l_{22}$$

$$P_{A36} = \bar{a}l_{36} + \bar{b}l_{35} + \bar{c}l_{34}$$

$$Q_{H29} = \bar{k}l_{30} + \bar{p}l_{29} + \bar{q}l_{28} + \bar{q}l_{32} + \bar{r}l_{31}$$

for a typical quadrant as illustrated in FIGS. 18A, 18B, 19A, 19B and 20A-20C, where P subscript and Q subscript represent the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated P and Q detector, respectively, illustrated in FIGS. 18A, 18B, 19A, 19B and 20A-20C, I zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{d} = C \int_0^{V_{1'PZONE}} \frac{e^{-\mu t_{1'PZONE}}}{l_{1'PZONE}^2} f_{C1'PZONE} dV$$

$$\bar{e} = C \int_0^{V_{2'PZONE}} \frac{e^{-\mu t_{2'PZONE}}}{l_{2'PZONE}^2} f_{C2'PZONE} dV$$

$$\bar{f} = C \int_0^{V_{2''PZONE}} \frac{e^{-\mu t_{2''PZONE}}}{l_{2''PZONE}^2} f_{C2''PZONE} dV$$

$$\bar{g} = C \int_0^{V_{3'PZONE}} \frac{e^{-\mu t_{3'PZONE}}}{l_{3'PZONE}^2} f_{C3'PZONE} dV$$

$$\bar{h} = C \int_0^{V_{3''PZONE}} \frac{e^{-\mu t_{3''PZONE}}}{l_{3''PZONE}^2} f_{C3''PZONE} dV$$

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^2} f_{C1QZONE} dV$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^2} f_{C2QZONE} dV$$

$$\bar{a} = C \int_0^{V_{1PZONE}} \frac{e^{-\mu t_{1PZONE}}}{l_{1PZONE}^2} f_{C1PZONE} dV$$

$$\bar{b} = C \int_0^{V_{2PZONE}} \frac{e^{-\mu t_{2PZONE}}}{l_{2PZONE}^2} f_{C2PZONE} dV$$

$$\bar{c} = C \int_0^{V_{3PZONE}} \frac{e^{-\mu t_{3PZONE}}}{l_{3PZONE}^2} f_{C3PZONE} dV$$

$$\bar{k} = C \int_0^{V_{1''P/QZONE}} \frac{e^{-\mu t_{1''P/QZONE}}}{l_{1''P/QZONE}^2} f_{C1''P/QZONE} dV$$

$$\bar{p} = C \int_0^{V_{1'QZONE}} \frac{e^{-\mu t_{1'QZONE}}}{l_{1'QZONE}^2} f_{C1'QZONE} dV$$

$$\bar{q} = C \int_0^{V_{2'QZONE}} \frac{e^{-\mu t_{2'QZONE}}}{l_{2'QZONE}^2} f_{C2'QZONE} dV$$

$$\bar{r} = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^2} f_{C2''QZONE} dV$$

$$\bar{f}_c = C \int_0^{V_{2''QZONE}} \frac{e^{-\mu t_{2''QZONE}}}{l_{2''QZONE}^2} f_{C2''QZONE} dV$$

where $C = K A_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

59. A method in accordance with claim 42 wherein said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantifying step includes the further step of quantifying at least one of said resultant radiation emission quantities for an associated one of said effective fields of view by utilizing an associated detection means including a detector, said

associated resultant radiation emission quantity being quantified at least in accordance with the expressions

$$P_{G1} = \bar{a}I_1 + \bar{b}I_{10} + \bar{c}I_{19}$$

$$P_{G7} = \bar{a}I_7 + \bar{b}I_{18} + \bar{c}I_{25}$$

$$Q_{C28} = \bar{m}I_{28} + \bar{n}I_{31}$$

$$P_{F11} = \bar{k}I_9 + \bar{d}I_{11} + \bar{e}I_{14} + \bar{f}I_{20} + \bar{g}I_{23} + \bar{h}I_{26} + \bar{g}I_{32} + \bar{h}I_{35}$$

$$P_{F12} = \bar{k}I_3 + \bar{d}I_{12} + \bar{e}I_{15} + \bar{f}I_{21} + \bar{g}I_{24} + \bar{g}I_{27} + \bar{g}I_{33} + \bar{h}I_{36}$$

for a typical quadrant as illustrated in FIGS. 21A, 21B, 22A, 22B, 23A and 23B, where P subscript and Q subscript represent the detected resultant radiation emission resulting from the contributions of all the detection zones within the effective field of view of an associated P and Q detector, respectively, illustrated in FIGS. 21A, 21B, 22A, 22B, 23A and 23B, I zone represents the radiation concentration of radioisotope contained within a given zone,

$$\bar{a} = C \int_0^{V_{1PZONE}} \frac{e^{-\mu t_{1PZONE}}}{l_{1PZONE}^2} f_{C1PZONE} dV,$$

$$\bar{b} = C \int_0^{V_{2PZONE}} \frac{e^{-\mu t_{2PZONE}}}{l_{2PZONE}^2} f_{C2PZONE} dV,$$

$$\bar{c} = C \int_0^{V_{3PZONE}} \frac{e^{-\mu t_{3PZONE}}}{l_{3PZONE}^2} f_{C3PZONE} dV,$$

$$\bar{m} = C \int_0^{V_{1QZONE}} \frac{e^{-\mu t_{1QZONE}}}{l_{1QZONE}^2} f_{C1QZONE} dV,$$

$$\bar{n} = C \int_0^{V_{2QZONE}} \frac{e^{-\mu t_{2QZONE}}}{l_{2QZONE}^2} f_{C2QZONE} dV,$$

$$\bar{k} = C \int_0^{V_{1^{III}P/QZONE}} \frac{e^{-\mu t_{1^{III}P/QZONE}}}{l_{1^{III}P/QZONE}^2}$$

$$f_{C1^{III}P/QZONE} dV,$$

$$\bar{d} = C \int_0^{V_{1'PZONE}} \frac{e^{-\mu t_{1'PZONE}}}{l_{1'PZONE}^2} f_{C1'PZONE} dV,$$

$$\bar{e} = C \int_0^{V_{2'PZONE}} \frac{e^{-\mu t_{2'PZONE}}}{l_{2'PZONE}^2} f_{C2'PZONE} dV,$$

$$\bar{f} = C \int_0^{V_{2''PZONE}} \frac{e^{-\mu t_{2''PZONE}}}{l_{2''PZONE}^2} f_{C2''PZONE} dV,$$

$$\bar{g} = C \int_0^{V_{3'PZONE}} \frac{e^{-\mu t_{3'PZONE}}}{l_{3'PZONE}^2} f_{C3'PZONE} dV, \text{ and}$$

$$\bar{h} = C \int_0^{V_{3''PZONE}} \frac{e^{-\mu t_{3''PZONE}}}{l_{3''PZONE}^2} f_{C3''PZONE} dV,$$

where $C = KA_d \epsilon \delta s / V_T$, K represents the dimensional constant, A_d represents the area of the detector, ϵ represents the efficiency of the detection system, δ represents the fraction of radioisotope remaining since its original assay at the time of the data collection for the local cerebral blood flow determination, s represents the attenuation of radiation by the scalp and skull, V_T represents the volume of the zone, μ represents the total attenuation constant for gamma radiation of a specific energy, t represents the total thickness of brain tissue through which the radiation is attenuated, l represents the distance from the center of the particular zone to the detector along the central axis of view, and f_c represents the collimator factor for a given zone.

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