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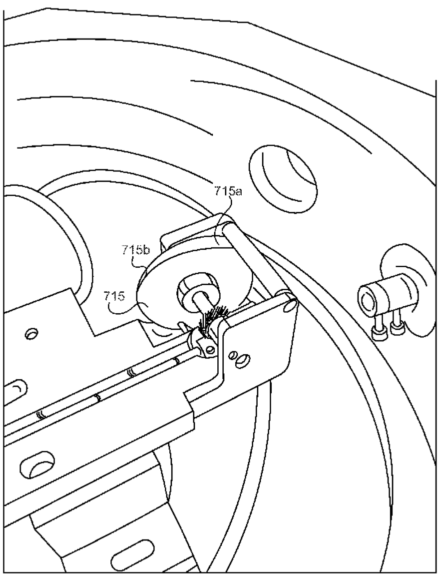

FIG. 18

(57) Abstract: An example particle accelerator includes a coil to provide a magnetic field to a cavity; a particle source to provide a plasma column to the cavity; a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, where the magnetic field causes particles accelerated from the plasma column to move orbitally within the cavity; an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and a structure arranged proximate to the extraction channel to change an energy level of the received particles.

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ADJUSTING ENERGY OF A PARTICLE BEAM

CROSS-REFERENCE TO RELATED APPLICATION

[0001] Priority is hereby claimed to U.S. Provisional Application No. 61/707,515, which was filed on September 28, 2012. The contents of U.S. Provisional Application No. 61/707,515 are hereby incorporated by reference into this disclosure.

TECHNICAL FIELD

[0002] This disclosure relates generally to adjusting energy of a particle beam, such as a proton or ion beam used in a particle therapy system.

BACKGROUND

[0003] Particle therapy systems use an accelerator to generate a particle beam for treating afflictions, such as tumors. In operation, the particle beam is accelerated inside a cavity of the particle accelerator, and removed from the cavity through an extraction channel. To track the extraction channel, the particle beam must be properly energized. Not enough energy, and the particle beam can collide with the inner edge of the extraction channel. Too much energy, and the particle beam can collide with the outer edge of the extraction channel. The result is that the particle beam is prevented from escaping the extraction channel, or the portion of the particle beam that does escape is compromised, thereby reducing treatment effectiveness.

[0004] Movement of the particle accelerator can affect the amount of energy in the particle beam that is received at the extraction channel.

SUMMARY

[0005] An example particle accelerator includes a coil to provide a magnetic field to a cavity; a particle source to provide a plasma column to the cavity; a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, where the magnetic field causes particles accelerated from the plasma column to move orbitally within the cavity; an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and a structure arranged proximate to the extraction channel to change an energy level of the received particles. This example particle accelerator may include one or more of the following features, either alone or in combination.

[0006] The structure may have multiple thicknesses. The structure may have variable thickness ranging from a maximum thickness to a minimum thickness. The structure may be movable relative to the extraction channel to place one of the multiple thicknesses in a path of the received particles. The structure may be wheel-shaped and may be rotatable within the extraction channel. The structure may include at least one of the following materials: beryllium, carbon and plastic.

[0007] The particle accelerator may be rotatable relative to a fixed position. The particle accelerator may include a control system to control movement of the structure based on a rotational position of the particle accelerator.

[0008] The particle accelerator may include a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel.

[0009] An example proton therapy system may include the foregoing particle accelerator, where the particles comprise protons; and a gantry on which the particle accelerator is mounted. The gantry is rotatable relative to a patient position. Protons are output essentially directly from the particle accelerator to the patient position.

[0010] An example particle accelerator includes a coil to provide a magnetic field to a cavity; a particle source to provide a plasma column to the cavity; a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, where the magnetic field causes particles accelerated from the plasma column to move orbitally within the cavity; an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel. The regenerator is movable within the cavity relative to orbits of the particles. This example particle accelerator may include one or more of the following features, either alone or in combination.

[0011] The regenerator may be configured to move radially relative to an approximate center of the cavity. An actuator may be configured to move the regenerator in response to a control signal. The particle accelerator may be rotatable relative to a fixed position. The particle accelerator may include a control system to generate the control signal to control movement of the regenerator based on a rotational position of the particle accelerator. The regenerator may include a ferromagnetic material, such as iron.

[0012] An example proton therapy system may include the foregoing particle accelerator, where the particles comprise protons; and a gantry on which the particle accelerator is mounted. The gantry may be rotatable relative to a patient position. Protons are output essentially directly from the particle accelerator to the patient position.

[0013] An example particle accelerator includes a coil to provide a magnetic field to a cavity; a particle source to provide a plasma column to the cavity; a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, where the magnetic field causes particles accelerated from the plasma column to move orbitally within the cavity; an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel. The enclosure includes magnetic structures, where at least one of the magnetic structures has a slot therein, where the slot contains a magnetic shim that is ferromagnetic and movable within the slot, where the magnetic shim is movable relative to the regenerator to affect an amount by which the regenerator adjusts the magnetic field. This example particle accelerator may include one or more of the following features, either alone or in combination.

[0014] The at least magnetic structure may have multiple slots therein. Each slot may contain a magnetic shim that is ferromagnetic and that is movable within the slot. Each magnetic shim may be movable relative to the regenerator to affect an amount by which the regenerator adjusts the magnetic field.

[0015] The particle accelerator may be rotatable relative to a fixed position. The particle accelerator may include a control system to generate a control signal to control movement of the magnetic shim (or multiple magnetic shims) based on a rotational position of the particle accelerator. The magnetic shim (or multiple magnetic shims) may be or include an electromagnet.

[0016] An example proton therapy system may include the foregoing particle accelerator, where the particles comprise protons; and a gantry on which the particle accelerator is mounted. The gantry is rotatable relative to a patient position. Protons are output essentially directly from the particle accelerator to the patient position.

[0017] An example particle accelerator may include a cryostat comprising a superconducting coil, where the superconducting coil conducts a current that generates a magnetic field; magnetic structures adjacent to the cryostat, where the cryostat is attached to the magnetic structures and the magnetic structures contain a cavity; a particle source to provide a plasma column to the cavity; a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, where the magnetic field cause particles accelerated from the plasma column to move orbitally within the cavity; an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and an actuator that is controllable to move the cryostat relative to the magnetic structures. This example particle accelerator may include one or more of the following features, either alone or in combination.

[0018] The particle accelerator may be rotatable relative to a fixed position. The particle accelerator may include a control system to generate a control signal to control the actuator based on a rotational position of the particle accelerator. The actuator may be controlled to control movement of the cryostat so as to compensate for effects of gravity on the superconducting coil.

[0019] An example proton therapy system may include the foregoing particle accelerator, where the particles comprise protons; and a gantry on which the particle accelerator is mounted. The gantry is rotatable relative to a patient position. Protons are output essentially directly from the particle accelerator to the patient position.

[0020] An example variable-energy particle accelerator includes: magnetic structures defining a cavity in which particles are accelerated for output as a particle beam that has a selected energy from among a range of energies; an extraction channel to receive the particle beam; and a structure proximate to the extraction channel to intercept the particle beam prior to the particle beam entering the extraction channel, where the structure is movable based on the selected energy, and where the structure is for absorbing at least some energy of the particle beam prior to the particle beam entering the extraction channel. The example variable-energy particle accelerator may include one or more of the following features, either alone or in combination.

[0021] The structure may be a wheel having varying thickness, where different thicknesses are capable of absorbing different amounts of energy. The variable-energy particle accelerator may include a magnetic regenerator to implement a magnetic field bump at a particle orbit that corresponds to the selected energy. The magnetic regenerator may be movable based on movement of the variable-energy particle accelerator. The magnetic regenerator may be movable to intercept a particle orbit having the selected energy.

[0022] Two or more of the features described in this disclosure, including those described in this summary section, may be combined to form implementations not specifically described herein.

[0023] Control of the various systems described herein, or portions thereof, may be implemented via a computer program product that includes instructions that are stored on one or more non-transitory machine-readable storage media, and that are executable on one or more processing devices. The systems described herein, or portions thereof, may be implemented as an apparatus, method, or electronic system that may include one or more processing devices and memory to store executable instructions to implement control of the stated functions.

[0024] The details of one or more implementations are set forth in the accompanying drawings and the description below. Other features, objects, and advantages will be apparent from the description and drawings, and from the claims.

DESCRIPTION OF DRAWINGS

[0025] Fig. 1 is a perspective view of an example therapy system.

[0026] Fig. 2 is an exploded perspective view of components of an example synchrocyclotron.

[0027] Figs. 3, 4, and 5 are cross-sectional views of an example synchrocyclotron.

[0028] Fig. 6 is a perspective view of an example synchrocyclotron.

[0029] Fig. 7 is a cross-sectional view of a portion of an example reverse bobbin and windings.

[0030] Fig. 8 is a cross sectional view of an example cable-in-channel composite conductor.

[0031] Fig. 9 is a cross-sectional view of an example ion source.

[0032] Fig. 10 is a perspective view of an example dee plate and an example dummy dee.

[0033] Fig. 11 is a perspective view of an example vault.

[0034] Fig. 12 is a perspective view of an example treatment room with a vault.

[0035] Fig. 13 shows an example of a patient relative to an accelerator.

[0036] Fig. 14 shows a patient positioned within an example inner gantry in a treatment room.

[0037] Fig. 15 is a top view of an example acceleration cavity and extraction channel.

[0038] Fig. 16 is a graph showing magnetic field strength versus radial distance from a plasma column, along with a cross-section of an example part of a cryostat of a superconducting magnet.

[0039] Fig. 17 is a top view of an example acceleration cavity and extraction channel, which depicts orbits moving to enter the extraction channel.

[0040] Fig. 18 is a perspective view of an example structure used to change the energy of a particle beam in the extraction channel.

[0041] Fig. 18A is a side view of the structure of Fig. 18.

[0042] Figs. 19 to 21 are top views of an example acceleration cavity and extraction channel, which depict moving the regenerator to primarily impact certain orbits of particles in the cavity.

[0043] Fig. 22 is a perspective view of an example magnetic shim.

[0044] Fig. 23 is cut-away side view of magnetic yokes, an acceleration cavity and a cold mass, which includes magnetic shims.

[0045] Fig. 24 is a cut-away perspective view of an example part of a cryostat.

[0046] Fig. 25 is a conceptual view of an example particle therapy system that may use a variable-energy particle accelerator.

[0047] Fig. 26 is an example graph showing energy and current for variations in magnetic field and distance in a particle accelerator.

[0048] Fig. 27 is a side view of an example structure for sweeping voltage on a dee plate over a frequency range for each energy level of a particle beam, and for varying the frequency range when the particle beam energy is varied.

[0049] Fig. 28 is a perspective, exploded view of an example magnet system that may be used in a variable-energy particle accelerator.

[0050] Like reference symbols in the various drawings indicate like elements.

DETAILED DESCRIPTION

OVERVIEW

[0051] Described herein is an example of a particle accelerator for use in a system, such as a proton or ion therapy system. The system includes a particle accelerator – in this example, a synchrocyclotron – mounted on a gantry. The gantry enables the accelerator to be rotated around a patient position, as explained in more detail below. In some implementations, the gantry is steel and has two legs mounted for rotation on two respective bearings that lie on opposite sides of a patient. The particle accelerator is supported by a steel truss that is long enough to span a treatment area in which the patient lies and that is attached stably at both ends to the rotating legs of the gantry. As a result of rotation of the gantry around the patient, the particle accelerator also rotates.

[0052] In an example implementation, the particle accelerator (e.g., the synchrocyclotron) includes a cryostat that holds a superconducting coil for conducting a current that generates a magnetic field (B). In this example, the cryostat uses liquid helium (He) to maintain the coil at superconducting temperatures, e.g., 4° Kelvin (K). Magnetic yokes are adjacent (e.g., around) the cryostat, and define a cavity in which particles are accelerated. The cryostat is attached to the magnetic yokes through straps or the like. While this attachment, and the attachment of the superconducting coil inside the cryostat, restricts movement of the superconducting coil, coil movement is not entirely prevented. For example, in some implementations, as a result of gravitational pull during rotation of the gantry, the superconducting coil is movable by small amounts (e.g., tenths of millimeters in some cases). As described below, this movement can affect the amount of energy in a particle beam that is received at an extraction channel and thereby affect the output of the particle accelerator.

[0053] In this example implementation, the particle accelerator includes a particle source (e.g., a Penning Ion Gauge – PIG source) to provide a plasma column to the cavity. Hydrogen gas is ionized to produce the plasma column. A voltage source provides a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column. As noted, in this example, the particle accelerator is a synchrocyclotron. Accordingly, the RF voltage is swept across a range of frequencies to account for relativistic effects on the particles (e.g., increasing particle mass) when extracting particles from the column. The magnetic field produced by the coil causes particles accelerated from the plasma column to accelerate orbitally within the cavity. A magnetic field regenerator is positioned in the cavity to adjust the existing magnetic field inside the cavity to thereby change locations of successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to an extraction channel that passes through the yokes. The regenerator may increase the magnetic field at a point in the cavity (e.g., it may produce a magnetic field "bump" at an area of the cavity), thereby causing each successive orbit of particles at that point to precess outward toward the entry point of the extraction channel, eventually reaching the extraction channel. The extraction channel receives particles accelerated from the plasma column and outputs the received particles from the cavity.

[0054] Movement of the superconducting coil can affect the locations of the orbits inside the cavity. For example, movement in one direction can cause lower-energy orbits to impact the regenerator, while movement in another direction can cause higher-energy orbits to impact the regenerator (particle orbit energy is proportional to the radial distance from the originating plasma column). So, in a case where overly low-energy orbits impact the regenerator, the particle beam may collide with the inner edge of the extraction channel, as noted above. In a case where overly high-energy orbits impact the regenerator, the particle beam may collide with the outer edge of the extraction channel, as noted above. The example systems described herein use techniques to compensate for these effects resulting from motion of the superconducting coil due to its rotation (e.g., due to the effect of gravity). A summary of these techniques is provided below, followed by a description of an example particle therapy system in which they may be implemented and more detailed descriptions of these various techniques.

[0055] In an example technique, a structure is incorporated proximate to (e.g., at the entry to or inside of) the extraction channel. The structure may be a rotatable variable-thickness wedge having a wheel-like shape. The structure absorbs energy of the particle beam, thereby allowing a lower-energy (e.g., appropriately energized) beam to pass through the extraction channel. The thicker portions of the structure absorb more energy than the thinner portions of the structure. In some implementations, the structure may contain no material at a point where the particle beam is meant to pass without any energy absorption. Alternatively, the structure may be movable out of the beam path. The structure thus enables the amount of energy in the beam to be variably adjusted. In some implementations, the structure is controlled based on a rotational position of the particle accelerator. For example, the position of the gantry may be determined, and that position may be used to control the rotational position of the energy-absorbing structure. Ideally, the structure minimizes scattering of the beam; however, in practice, there may be amounts of scatter that are present and that are tolerable.

[0056] In another example technique, the physical position of the regenerator within the cavity may be adjustable to compensate for movement of the superconducting coil. For example, computer-controlled actuators may be used to adjust the position of the regenerator within the cavity based, e.g., on a rotational position of the particle accelerator. By so adjusting the position of the regenerator, it may be possible to position the regenerator so that the appropriate adjustment to the magnetic field resulting from the regenerator impacts the proper particle orbits regardless of the rotational position of the particle accelerator.

[0057] The regenerator is typically made of a ferromagnetic material. It is therefore possible to adjust the magnetic strength of the regenerator using one or more magnetic shims. Accordingly, in another example technique, it is possible to adjust the magnetic field of the regenerator (e.g., to increase or decrease the magnetic field bump produced by the regenerator) or to move the effective location of the magnetic field perturbation produced by the regenerator without physically moving the regenerator. For example, if movement of the superconducting coil results in lower-energy orbits impacting the regenerator, the magnetic field of the regenerator can be decreased so that it doesn't begin to perturb beam orbits until higher energy orbits reach it. It could also be effectively moved radially outward while maintaining the same overall strength (peak field) so that the orbits gain higher energy before being effected by the regenerator. Likewise if the superconducting coil movement results in higher energy orbits impacting the regenerator the strength of the regenerator can be increased or positioned radially inward to interact with orbits at lower energies. In an example implementation, the magnetic field is adjusted by moving a magnetic shim (e.g., a metal plunger) within a slot/hole in a magnetic yoke that is near to the regenerator. The magnetic shim is made of ferromagnetic material and its proximity to the regenerator affects the magnetic field of the regenerator. Moving the magnetic shim closer to the regenerator (e.g., further inside the slot increases the magnetic field produced by the regenerator; and moving the magnetic shim away from the regenerator (e.g., upwards in, or outside of, the slot) decreases the magnetic field produced by the regenerator. In another example the magnetic shim can be placed radially closer to the center of the cyclotron than the regenerator magnetic center. When the shim is place closer to the acceleration plane, it moves the effective center of the regenerator magnetic perturbation without appreciably changing the peak magnetic field strength. The magnetic shim may be computer-controlled to vary its position based, e.g., on a rotational position of the particle accelerator.

[0058] In some implementations, more than one magnetic shim may be used. In still other implementations, miniature electromagnet(s) may be used as a magnetic shim, and the current therethrough controlled based, e.g., on a rotational position of the particle accelerator.

[0059] In another example, the entire cryostat may be moved relative to the yokes to compensate for movement of the superconducting coil. For example, movement of the cryostat can affect which orbits of particles impact the regenerator. So, if movement of the superconducting coil occurs in one direction, the cryostat may be moved in the direction to compensate for that movement and cause the superconducting coil to be properly repositioned.

[0060] The foregoing techniques for adjusting the energy of a particle beam in a particle accelerator may be used individually in a single particle accelerator, or any two or more of those techniques may be used in any appropriate combination in a single particle accelerator. An example of a particle therapy system in which the foregoing techniques may be used is provided below.

EXAMPLE PARTICLE THERAPY SYSTEM

[0061] Referring to Fig 1, a charged particle radiation therapy system 500 includes a beam-producing particle accelerator 502 having a weight and size small enough to permit it to be mounted on a rotating gantry 504 with its output directed straight (that is, essentially directly) from the accelerator housing toward a patient 506.

[0062] In some implementations, the steel gantry has two legs 508, 510 mounted for rotation on two respective bearings 512, 514 that lie on opposite sides of the patient. The accelerator is supported by a steel truss 516 that is long enough to span a treatment area 518 in which the patient lies (e.g., twice as long as a tall person, to permit the person to be rotated fully within the space with any desired target area of the patient remaining in the line of the beam) and is attached stably at both ends to the rotating legs of the gantry.

[0063] In some examples, the rotation of the gantry is limited to a range 520 of less than 360 degrees, e.g., about 180 degrees, to permit a floor 522 to extend from a wall of the vault 524 that houses the therapy system into the patient treatment area. The limited rotation range of the gantry also reduces the required thickness of some of the walls, which provide radiation shielding of people outside the treatment area. A range of 180 degrees of gantry rotation is enough to cover all treatment approach angles, but providing a larger range of travel can be useful. For example the range of rotation may be between 180 and 330 degrees and still provide clearance for the therapy floor space.

[0064] The horizontal rotational axis 532 of the gantry is located nominally one meter above the floor where the patient and therapist interact with the therapy system. This floor is positioned about 3 meters above the bottom floor of the therapy system shielded vault. The accelerator can swing under the raised floor for delivery of treatment beams from below the rotational axis. The patient couch moves and rotates in a substantially horizontal plane parallel to the rotational axis of the gantry. The couch can rotate through a range 534 of about 270 degrees in the horizontal plane with this configuration. This combination of gantry and patient rotational ranges and degrees of freedom allow the therapist to select virtually any approach angle for the beam. If needed, the patient can be placed on the couch in the opposite orientation and then all possible angles can be used.

[0065] In some implementations, the accelerator uses a synchrocyclotron configuration having a very high magnetic field superconducting electromagnetic structure. Because the bend radius of a charged particle of a given kinetic energy is reduced in direct proportion to an increase in the magnetic field applied to it, the very high magnetic field superconducting magnetic structure permits the accelerator to be made smaller and lighter. The synchrocyclotron uses a magnetic field that is uniform in rotation angle and falls off in strength with increasing radius. Such a field shape can be achieved regardless of the magnitude of the magnetic field, so in theory there is no upper limit to the magnetic field strength (and therefore the resulting particle energy at a fixed radius) that can be used in a synchrocyclotron.

[0066] Superconducting materials lose their superconducting properties in the presence of very high magnetic fields. High performance superconducting wire windings are used to allow very high magnetic fields to be achieved.

[0067] Superconducting materials typically need to be cooled to low temperatures for their superconducting properties to be realized. In some examples described here, cryo-coolers are used to bring the superconducting coil windings to temperatures near absolute zero. Using cryo-coolers can reduce complexity and cost.

[0068] The synchrocyclotron is supported on the gantry so that the beam is generated directly in line with the patient. The gantry permits rotation of the cyclotron about a horizontal rotational axis that contains a point (isocenter 540) within, or near, the patient. The split truss that is parallel to the rotational axis, supports the cyclotron on both sides.

[0069] Because the rotational range of the gantry is limited, a patient support area can be accommodated in a wide area around the isocenter. Because the floor can be extended broadly around the isocenter, a patient support table can be positioned to move relative to and to rotate about a vertical axis 542 through the isocenter so that, by a combination of gantry rotation and table motion and rotation, any angle of beam direction into any part of the patient can be achieved. The two gantry arms are separated by more than twice the height of a tall patient, allowing the couch with patient to rotate and translate in a horizontal plane above the raised floor.

[0070] Limiting the gantry rotation angle allows for a reduction in the thickness of at least one of the walls surrounding the treatment room. Thick walls, typically constructed of concrete, provide radiation protection to individuals outside the treatment room. A wall downstream of a stopping proton beam may be about twice as thick as a wall at the opposite end of the room to provide an equivalent level of protection. Limiting the range of gantry rotation enables the treatment room to be sited below earth grade on three sides, while allowing an occupied area adjacent to the thinnest wall reducing the cost of constructing the treatment room.

[0071] In the example implementation shown in Fig. 1, the superconducting synchrocyclotron 502 operates with a peak magnetic field in a pole gap of the synchrocyclotron of 8.8 Tesla. The synchrocyclotron produces a beam of protons having an energy of 250 MeV. In other implementations the field strength could be in the range of 4 to 20 Tesla or 6 to 20 Tesla and the proton energy could be in the range of 150 to 300 MeV

[0072] The radiation therapy system described in this example is used for proton radiation therapy, but the same principles and details can be applied in analogous systems for use in heavy ion (ion) treatment systems.

[0073] As shown in Figs. 2, 3, 4, 5, and 6, an example synchrocyclotron 10 (e.g., 502 in Fig. 1) includes a magnet system 12 that contains an particle source 90, a radiofrequency drive system 91, and a beam extraction system 38. The magnetic field established by the magnet system has a shape appropriate to maintain focus of a contained proton beam using a combination of a split pair of annular superconducting coils 40, 42 and a pair of shaped ferromagnetic (e.g., low carbon steel) pole faces 44, 46.

[0074] The two superconducting magnet coils are centered on a common axis 47 and are spaced apart along the axis. As shown in Figs. 7 and 8, the coils are formed by of Nb_3Sn-based superconducting 0.8 mm diameter strands 48 (that initially comprise a niobium-tin core surrounded by a copper sheath) deployed in a twisted cable-in-channel conductor geometry. After seven individual strands are cabled together, they are heated to cause a reaction that forms the final (brittle) superconducting material of the wire. After the material has been reacted, the wires are soldered into the copper channel (outer dimensions 3.18 x 2.54 mm and inner dimensions 2.08 x 2.08 mm) and covered with insulation 52 (in this example, a woven fiberglass material). The copper channel containing the wires 53 is then wound in a coil having a rectangular cross-section of 8.55 cm x 19.02 cm, having 26 layers and 49 turns per layer. The wound coil is then vacuum impregnated with an epoxy compound. The finished coils are mounted on an annular stainless steel reverse bobbin 56. Heater blankets 55 are placed at intervals in the layers of the windings to protect the assembly in the event of a magnet quench.

[0075] The entire coil can then be covered with copper sheets to provide thermal conductivity and mechanical stability and then contained in an additional layer of epoxy. The precompression of the coil can be provided by heating the stainless steel reverse bobbin and fitting the coils within the reverse bobbin. The reverse bobbin inner diameter is chosen so that when the entire mass is cooled to 4 K, the reverse bobbin stays in contact with the coil and provides some compression. Heating the stainless steel reverse bobbin to approximately 50 degrees C and fitting coils at a temperature of 100 degrees Kelvin can achieve this.

[0076] The geometry of the coil is maintained by mounting the coils in a reverse rectangular bobbin 56 to exert a restorative force 60 that works against the distorting force produced when the coils are energized. As shown in Fig. 5, the coil position is maintained relative to the magnet yoke and cryostat using a set of warm-to-cold support straps 402, 404, 406. Supporting the cold mass with thin straps reduces the heat leakage imparted to the cold mass by the rigid support system.

The straps are arranged to withstand the varying gravitational force on the coil as the magnet rotates on board the gantry. They withstand the combined effects of gravity and the large de-centering force realized by the coil when it is perturbed from a perfectly symmetric position relative to the magnet yoke. Additionally the links act to reduce dynamic forces imparted on the coil as the gantry accelerates and decelerates when its position is changed. Each warm-to-cold support includes one S2 fiberglass link and one carbon fiber link. The carbon fiber link is supported across pins between the warm yoke and an intermediate temperature (50 - 70 K), and the S2 fiberglass link 408 is supported across the intermediate temperature pin and a pin attached to the cold mass. Each link is 5 cm long (pin center to pin center) and is 17 mm wide. The link thickness is 9 mm. Each pin is made of high strength stainless steel and is 40 mm in diameter.

[0077] Referring to Fig. 3, the field strength profile as a function of radius is determined largely by choice of coil geometry and pole face shape; the pole faces 44, 46 of the permeable yoke material can be contoured to fine tune the shape of the magnetic field to ensure that the particle beam remains focused during acceleration.

[0078] The superconducting coils are maintained at temperatures near absolute zero (e.g., about 4 degrees Kelvin) by enclosing the coil assembly (the coils and the bobbin) inside an evacuated annular aluminum or stainless steel cryostatic chamber 70 that provides a free space around the coil structure, except at a limited set of support points 71, 73. In an alternate version (Fig. 4) the outer wall of the cryostat may be made of low carbon steel to provide an additional return flux path for the magnetic field.

[0079] In some implementations, the temperature near absolute zero is achieved and maintained using one single-stage Gifford-McMahon cryo-cooler and three two-stage Gifford McMahon cryo-coolers. Each two stage cryo-cooler has a second stage cold end attached to a condenser that recondenses Helium vapor into liquid Helium. The cryo-cooler heads are supplied with compressed Helium from a compressor. The single-stage Gifford-McMahon cryo-cooler is arranged to cool high temperature (e.g., 50 - 70 degrees Kelvin) leads that supply current to the superconducting windings.

[0080] In some implementations, the temperature near absolute zero is achieved and maintained using two Gifford-McMahon cryo-coolers 72, 74 that are arranged at different positions on the coil assembly. Each cryo-cooler has a cold end 76 in contact with the coil assembly. The cryo-cooler heads 78 are supplied with compressed Helium from a compressor 80. Two other Gifford-McMahon cryo-coolers 77, 79 are arranged to cool high temperature (e.g., 60 - 80 degrees Kelvin) leads that supply current to the superconducting windings.

[0081] The coil assembly and cryostatic chambers are mounted within and fully enclosed by two halves 81, 83 of a pillbox-shaped magnet yoke 82. In this example, the inner diameter of the coil assembly is about 74.6 cm. The iron yoke 82 provides a path for the return magnetic field flux 84 and magnetically shields the volume 86 between the pole faces 44, 46 to prevent external magnetic influences from perturbing the shape of the magnetic field within that volume. The yoke also serves to decrease the stray magnetic field in the vicinity of the accelerator. In some implementations, the synchrocyclotron may have an active return system to reduce stray magnetic fields. An example of an active return system is described in U.S.

Patent Application No. 13/907,601, which was filed on May 31, 2013, the contents of which are incorporated herein by reference. In the active return system, the relatively large magnetic yokes described herein are replaced by smaller magnetic structures, referred to as pole pieces. Superconducting coils run current opposite to the main coils described herein in order to provide magnetic return and thereby reduce stray magnetic fields

[0082] As shown in Figs. 3 and 9, the synchrocyclotron includes a particle source 90 of a Penning ion gauge geometry located near the geometric center 92 of the magnet structure 82. The particle source may be as described below, or the particle source may be of the type described in U.S. Patent Application No. 11/948,662 incorporated herein by reference.

[0083] Particle source 90 is fed from a supply 99 of hydrogen through a gas line 101 and tube 194 that delivers gaseous hydrogen. Electric cables 94 carry an electric current from a current source 95 to stimulate electron discharge from cathodes 192, 190 that are aligned with the magnetic field, 200.

[0084] In some implementations, the gas in gas tube 101 may include a mixture of hydrogen and one or more other gases. For example, the mixture may contain hydrogen and one or more of the noble gases, e.g., helium, neon, argon, krypton, xenon and/or radon (although the mixture is not limited to use with the noble gases). In some implementations, the mixture may be a mixture of hydrogen and helium. For example, the mixture may contain about 75% or more of hydrogen and about 25% or less of helium (with possible trace gases included). In another example, the mixture may contain about 90% or more of hydrogen and about 10% or less of helium (with possible trace gases included). In examples, the hydrogen/helium mixture may be any of the following: >95%/<5%, >90%/<10%, >85%/<15%, >80%/<20%, >75%/<20%, and so forth.

[0085] Possible advantages of using a noble (or other) gas in combination with hydrogen in the particle source may include: increased beam intensity, increased cathode longevity, and increased consistency of beam output.

[0086] In this example, the discharged electrons ionize the gas exiting through a small hole from tube 194 to create a supply of positive ions (protons) for acceleration by one semicircular (dee-shaped) radio-frequency plate 100 that spans half of the space enclosed by the magnet structure and one dummy dee plate 102. In the case of an interrupted particle source (an example of which is described in U.S. Patent Application No. 11/948,662), all (or a substantial part) of the tube containing plasma is removed at the acceleration region, thereby allowing ions to be more rapidly accelerated in a relatively high magnetic field.

[0087] As shown in Fig. 10, the dee plate 100 is a hollow metal structure that has two semicircular surfaces 103, 105 that enclose a space 107 in which the protons are accelerated during half of their rotation around the space enclosed by the magnet structure. A duct 109 opening into the space 107 extends through the yoke to an external location from which a vacuum pump 111 can be attached to evacuate the space 107 and the rest of the space within a vacuum chamber 119 in which the acceleration takes place. The dummy dee 102 comprises a rectangular metal ring that is spaced near to the exposed rim of the dee plate. The dummy dee is grounded to the vacuum chamber and magnet yoke. The dee plate 100 is driven by a radio-frequency signal that is applied at the end of a radio-frequency transmission line to impart an electric field in the space 107. The radio frequency electric field is made to vary in time as the accelerated particle beam increases in distance from the geometric center. The radio frequency electric field may be controlled in the manner described in U.S. Patent Application No. 11/948,359, entitled "Matching A Resonant Frequency Of A Resonant Cavity To A Frequency Of An Input Voltage", the contents of which are incorporated herein by reference.

[0088] For the beam emerging from the centrally located particle source to clear the particle source structure as it begins to spiral outward, a large voltage difference is required across the radio frequency plates. 20,000 Volts is applied across the radio frequency plates. In some versions from 8,000 to 20,000 Volts may be applied across the radio frequency plates. To reduce the power required to drive this large voltage, the magnet structure is arranged to reduce the capacitance between the radio frequency plates and ground. This is done by forming holes with sufficient clearance from the radio frequency structures through the outer yoke and the cryostat housing and making sufficient space between the magnet pole faces.

[0089] The high voltage alternating potential that drives the dee plate has a frequency that is swept downward during the accelerating cycle to account for the increasing relativistic mass of the protons and the decreasing magnetic field. The dummy dee does not require a hollow semi-cylindrical structure as it is at ground potential along with the vacuum chamber walls. Other plate arrangements could be used such as more than one pair of accelerating electrodes driven with different electrical phases or multiples of the fundamental frequency. The RF structure can be tuned to keep the Q high during the required frequency sweep by using, for example, a rotating capacitor having intermeshing rotating and stationary blades. During each meshing of the blades, the capacitance increases, thus lowering the resonant frequency of the RF structure. The blades can be shaped to create a precise frequency sweep required. A drive motor for the rotating condenser can be phase locked to the RF generator for precise control. One bunch of particles is accelerated during each meshing of the blades of the rotating condenser.

[0090] The vacuum chamber 119 in which the acceleration occurs is a generally cylindrical container that is thinner in the center and thicker at the rim. The vacuum chamber encloses the RF plates and the particle source and is evacuated by the vacuum pump 111. Maintaining a high vacuum insures that accelerating ions are not lost to collisions with gas molecules and enables the RF voltage to be kept at a higher level without arcing to ground.

[0091] Protons traverse a generally spiral orbital path beginning at the particle source. In half of each loop of the spiral path, the protons gain energy as they pass through the RF electric field in space 107. As the ions gain energy, the radius of the central orbit of each successive loop of their spiral path is larger than the prior loop until the loop radius reaches the maximum radius of the pole face. At that location a magnetic and electric field perturbation directs ions into an area where the magnetic field rapidly decreases, and the ions depart the area of the high magnetic field and are directed through an evacuated tube 38, referred to herein as the extraction channel, to exit the yoke of the cyclotron. A magnetic regenerator may be used to change the magnetic field perturbation to direct the ions. The ions exiting the cyclotron will tend to disperse as they enter the area of markedly decreased magnetic field that exists in the room around the cyclotron. Beam shaping elements 107, 109 in the extraction channel 38 redirect the ions so that they stay in a straight beam of limited spatial extent.

[0092] The magnetic field within the pole gap needs to have certain properties to maintain the beam within the evacuated chamber as it accelerates. The magnetic field index n, which is shown below,

$$n = -(r/B)dB/dr,$$

should be kept positive to maintain this "weak" focusing. Here r is the radius of the beam and B is the magnetic field. Additionally, in some implementations, the field index needs to be maintained below 0.2, because at this value the periodicity of radial oscillations and vertical oscillations of the beam coincide in a $\nu r = 2\ \nu_z$ resonance. The betatron frequencies are defined by $\nu_r = (1-n)^{1/2}$ and $\nu_z = n^{1/2}$. The ferromagnetic pole face is designed to shape the magnetic field generated by the coils so that the field index n is maintained positive and less than 0.2 in the smallest diameter consistent with a 250 MeV beam in the given magnetic field.

[0093] As the beam exits the extraction channel it is passed through a beam formation system 125 (Fig. 5) that can be programmably controlled to create a desired combination of scattering angle and range modulation for the beam. Beam formation system 125 may be used in conjunction with an inner gantry 601 (Fig. 14) to direct a beam to the patient.

[0094] During operation, the plates absorb energy from the applied radio frequency field as a result of conductive resistance along the surfaces of the plates. This energy appears as heat and is removed from the plates using water cooling lines 108 that release the heat in a heat exchanger 113 (Fig. 3).

[0095] Stray magnetic fields exiting from the cyclotron are limited by both the pillbox magnet yoke (which also serves as a shield) and a separate magnetic shield

114. The separate magnetic shield includes of a layer 117 of ferromagnetic material (e.g., steel or iron) that encloses the pillbox yoke, separated by a space 116. This configuration that includes a sandwich of a yoke, a space, and a shield achieves adequate shielding for a given leakage magnetic field at lower weight.

[0096] As mentioned, the gantry allows the synchrocyclotron to be rotated about the horizontal rotational axis 532. The truss structure 516 has two generally parallel spans 580, 582. The synchrocyclotron is cradled between the spans about midway between the legs. The gantry is balanced for rotation about the bearings using counterweights 122, 124 mounted on ends of the legs opposite the truss.

[0097] The gantry is driven to rotate by an electric motor mounted to one or both of the gantry legs and connected to the bearing housings by drive gears . The rotational position of the gantry is derived from signals provided by shaft angle encoders incorporated into the gantry drive motors and the drive gears.

[0098] At the location at which the ion beam exits the cyclotron, the beam formation system 125 acts on the ion beam to give it properties suitable for patient treatment. For example, the beam may be spread and its depth of penetration varied to provide uniform radiation across a given target volume. The beam formation system can include passive scattering elements as well as active scanning elements.

[0099] All of the active systems of the synchrocyclotron (the current driven superconducting coils, the RF-driven plates, the vacuum pumps for the vacuum acceleration chamber and for the superconducting coil cooling chamber, the current driven particle source, the hydrogen gas source, and the RF plate coolers, for example), may be controlled by appropriate synchrocyclotron control electronics (not shown), which may include, e.g., one or more computers programmed with appropriate programs to effect control.

[00100] The control of the gantry, the patient support, the active beam shaping elements, and the synchrocyclotron to perform a therapy session is achieved by appropriate therapy control electronics (not shown).

[00101] As shown in Figs. 1, 11, and 12, the gantry bearings are supported by the walls of a cyclotron vault 524. The gantry enables the cyclotron to be swung through a range 520 of 180 degrees (or more) including positions above, to the side of, and below the patient. The vault is tall enough to clear the gantry at the top and bottom extremes of its motion. A maze 146 sided by walls 148, 150 provides an entry and exit route for therapists and patients. Because at least one wall 152 is not in line with the proton beam directly from the cyclotron, it can be made relatively thin and still perform its shielding function. The other three side walls 154, 156, 150/148 of the room, which may need to be more heavily shielded, can be buried within an earthen hill (not shown). The required thickness of walls 154, 156, and 158 can be reduced, because the earth can itself provide some of the needed shielding.

[00102] Referring to Figs. 12 and 13, for safety and aesthetic reasons, a therapy room 160 may be constructed within the vault. The therapy room is cantilevered from walls 154, 156, 150 and the base 162 of the containing room into the space between the gantry legs in a manner that clears the swinging gantry and also maximizes the extent of the floor space 164 of the therapy room. Periodic servicing of the accelerator can be accomplished in the space below the raised floor. When the accelerator is rotated to the down position on the gantry, full access to the accelerator is possible in a space separate from the treatment area. Power supplies, cooling equipment, vacuum pumps and other support equipment can be located under the raised floor in this separate space. Within the treatment room, the patient support 170 can be mounted in a variety of ways that permit the support to be raised and lowered and the patient to be rotated and moved to a variety of positions and orientations.

[00103] In system 602 of Fig. 14, a beam-producing particle accelerator of the type described herein, in this case synchrocyclotron 604, is mounted on rotating gantry 605. Rotating gantry 605 is of the type described herein, and can angularly rotate around patient support 606. This feature enables synchrocyclotron 604 to provide a particle beam directly to the patient from various angles. For example, as in Fig. 14, if synchrocyclotron 604 is above patient support 606, the particle beam may be directed downwards toward the patient. Alternatively, if synchrocyclotron 604 is below patient support 606, the particle beam may be directed upwards toward the patient. The particle beam is applied directly to the patient in the sense that an intermediary beam routing mechanism is not required. A routing mechanism, in this context, is different from a shaping or sizing mechanism in that a shaping or sizing mechanism does not re-route the beam, but rather sizes and/or shapes the beam while maintaining the same general trajectory of the beam.

[00104] Further details regarding an example implementation of the foregoing system may be found in U.S. Patent No. 7,728,311, filed on November 16, 2006 and entitled “Charged Particle Radiation Therapy”, and in U.S. Patent Application No. 12/275,103, filed on November 20, 2008 and entitled “Inner Gantry”. The contents of U.S. Patent No. 7,728,311 and in U.S. Patent Application No. 12/275,103 are hereby incorporated by reference into this disclosure. In some implementations, the synchrocyclotron may be a variable-energy device, such as that described in U.S. Patent Application No. 13/916,401, filed on June 12, 2013, the contents of which are incorporated herein by reference.

EXAMPLE IMPLEMENTATIONS

[00105] Fig. 15 shows a top view of a portion of a cavity 700 in which particles are accelerated orbitally (e.g., in outward spiral orbits). A particle source 701, examples of which are described above, is disposed at about the center of the cavity. Charged particles (e.g., protons or ions) are extracted from a plasma column generated by particle source 701. The charged particles accelerate outwardly in orbits toward, and eventually reaching, magnetic regenerator 702. In this example implementation, regenerator 702 is a ferromagnetic structure made, e.g., of steel, iron, or any other type of ferromagnetic material. Regenerator 702 alters the background magnetic field that causes the outward orbital acceleration. In this example, regenerator 702 augments that magnetic field (e.g., it provides a bump in the field). The bump in the background magnetic field affects the particle orbits in a way that causes the orbits to move outwardly towards extraction channel 703. Eventually, the orbits enter extraction channel 703, from which they exit.

[00106] In more detail, a particle beam orbit approaches and interacts with regenerator 702. As a result of the increased magnetic field, the particle beam turns a bit more there and, instead of being circular, it precesses to the extraction channel. Fig. 16 shows the magnetic field (B) plotted against the radius (r) relative to the particle source 702. As shown in Fig. 16, in this example, B varies from about 9 Tesla (T) to about -2T. The 9T occurs at about the center of cavity 700. The polarity of the magnetic field changes after the magnetic field crosses the superconducting coil, resulting in about -2T on the exterior of the coil, eventually fading to about zero. The magnetic field bump 705 occurs at the point of the regenerator. Fig. 16 also shows the magnetic field plot relative to a cross-section 706 of a bobbin 706 having extraction channel 703 between two superconducting coils 709, 710.

[00107] Referring to Fig. 17, regenerator 702 causes changes in the angle and pitch of orbits 710 so that they move toward extraction channel 703. At the point of the extraction channel, the magnetic field strength is sufficiently low to enable the particle beam to enter the extraction channel and to proceed therethrough. Referring back to Fig. 15, extraction channel 703 contains various magnetic structures 711 for adding and/or subtracting dipole fields to direct the entering particle beam through extraction channel 703, to beam shaping elements.

[00108] In order to reach the exit point, the particle beam should have the appropriate amount of energy. The amount of energy required to reach that point may vary based, e.g., on the size of the accelerator and the length of the extraction channel (in this example, the extraction channel is about 1.7 or 2 meters in length). In this regard, at least part of extraction channel 703 is above the superconducting coil. As such, the magnetic field in the extraction channel may change little in response to accelerator rotation. Accordingly, the amount of energy needed for a particle beam to traverse the extraction channel may not change appreciably in response to the rotation of the particle accelerator.

[00109] As explained above, as the superconducting coil moves during rotation, orbits that are affected by regenerator 702 change due to gravitational movement of the coil. As noted, this movement can be as little as tenths of millimeters. Nevertheless, as a result, the energy of the particle beam that enters the extraction channel may be different from the energy required to traverse the entire channel. To adjust for this change in the energy of particles entering the extraction channel, a structure 715 may be placed inside, or at the entry point to, extraction channel 703. The structure may be used to absorb excess energy in the particle beam. In this example, structure 715 is a rotatable, variable-thickness wedge, which may have a wheel-like shape. An example of structure 715 is shown in Figs. 18 and 18A. As shown in these figures, structure 715 may have continuously varying thickness. Alternatively, the thicknesses may vary step-wise.

[00110] The structure may be moved (e.g., rotated) to absorb an appropriate amount of energy from a particle beam in/entering the extraction channel. In this implementation, thicker parts 715a of the structure absorb more energy than thinner parts 715b. Accordingly, the structure may be moved (e.g., rotated) to absorb different amounts of energy in a particle beam. In some implementations, the structure may have a part containing no material (e.g., a "zero" thickness), which allows the particle beam to pass unaltered. Alternatively, in such cases, the structure may be moved entirely or partly out of the beam path. In some implementations, the maximum thickness may be on the order of centimeters; however, the maximum thickness will vary from system-to-system based, e.g., on energy absorbing requirements. Fig. 18A also shows a motor 716 that controls an axle to rotate structure 715, e.g., in response to a detected gantry position.

[00111] The structure may be made of any appropriate material that is capable of absorbing energy in a particle beam. As noted above, ideally, the structure minimizes scattering of the particle beam in the extraction channel; however, in practice, there may be amounts of scatter that are present and that are tolerable. Examples of materials that may be used for the structure include, but are not limited to, beryllium, plastic containing hydrogen, and carbon. These materials may be used alone, in combination, or in combination with other materials.

[00112] The movement (e.g., rotation) of the structure may be computer-controlled using a control system that is part of the broader particle therapy system. Computer control may include generating one or more control signals to control movement of mechanical devices, such as actuators and motors that produce the motion. The rotation of structure 715 may be controlled based on a rotational position of the particle accelerator, as measured by the rotational position of the gantry (see, e.g., Figs. 1, 11 and 12 showing gantry rotation) on which the particle accelerator is mounted. The various parameters used to set the rotational position of the structure vis-à-vis the position of the gantry may be measured empirically, and programmed into the control system computer.

[00113] As noted above, in some implementations, the magnetic field in the extraction channel may change, albeit very little, in response to accelerator rotation. The amount of the change may be, e.g., a few tenths of a percent. In a specific example, this is reflected by a change of about six amperes (amps) of current out of a normal ~2000 amps running through the superconducting coil. This can affect the energy required for a particle beam to traverse the extraction channel. This small change in magnetic field may be adjusted by controlling the current through the superconducting coil or by controlling the rotation of structure 715.

[00114] In other implementations, adjusting the energy of particle beams reaching the extraction channel may be achieved by physically moving regenerator

702 so that, at different rotational positions, the regenerator affects different particle orbits. As above, the movement or regenerator 702 may be computer-controlled through a control system that is part of the particle therapy system. For example, the movement of regenerator 702 may be controlled based on a rotational position of the particle accelerator, as measured by the rotational position of the gantry on which the particle accelerator is mounted. The various parameters used to set the location of the regenerator vis-à-vis the rotational position of the gantry may be measured empirically, and programmed into the control system computer. One or more computer-controlled actuators may effect actual movement of the regenerator.

[00115] Referring to Fig. 19 for example, regenerator 702 may be positioned at location 717 initially, e.g., at a predefined initial position of the accelerator. In this position, the magnetic field bump produced by the regenerator has a primary impact on orbit 719 (to direct particles at that orbital position to the extraction channel). Orbit 720 is further from the location 721 of the plasma column than orbit 719. Consequently, orbit 720 has higher energy than orbit 719. Orbit 722 is closer to the location 721 of the plasma column than orbit 719. Consequently, orbit 722 has lower energy than orbit 719. As shown in Fig. 20, movement of the superconducting coil as a result of rotation can cause lower-energy orbit 722 to move into the path of regenerator 702 such that regenerator 702 primarily affects orbit 722. However, because orbit 722 is lower energy, it may not be capable of traversing the extraction channel and may impact the inner wall of the extraction channel before exiting. Accordingly, regenerator 702 may be moved from location 717 to location 723 (as depicted by arrow 724 of Fig. 21) so that regenerator 702 again primarily impacts orbit 719. The converse may be true as well. That is, if the superconducting coil moves such that an overly high-energy orbit 720 is primarily impacted by regenerator 702, regenerator 702 may be moved in the other direction (e.g., towards location 721) so that it primarily impacts the lower-energy orbit 719 (which has also moved). Although the figures depict movement of the regenerator in one dimension (radially), the regenerator may be moved in two or three dimensions, e.g., it may be moved in the Cartesian X, Y and/or Z directions.

[00116] In other implementations, the orbit that is affected primarily by the regenerator may be changed by altering the magnetic field (the magnetic field bump). This may be done, e.g., by changing the amount of ferromagnetic material in proximity to the regenerator. In an implementation, one or more magnetic shims may be used to alter the shape and/or strength of the magnetic field produced by the regenerator. In this regard, the regenerator may be made of a ferromagnetic material, such as steel (although other materials may be used in place of, or in addition to, steel). The magnetic shims may be a ferromagnetic material that is different from, or the same as, the material of which the regenerator is made.

[00117] In this implementation, the magnetic shims includes one or iron or steel magnetic shims. An example is magnetic shim 730 shown in Fig. 22; however, any appropriate shape may be used. For example, magnetic shim 730 may be in the shape of a rod or may have other appropriate shapes. Referring to Fig. 23, magnetic shims 730a, 730b may be placed in a slot of the corresponding yoke 731a, 731b near to the regenerator 702 or in the regenerator itself. Moving the magnetic shim downward, further inside a slot in the yoke increases the amount of ferromagnetic material near to the regenerator and thereby alters the location and size of the magnetic field bump produced by the regenerator. By contrast, moving a magnetic shim upward and out of the yoke decreases the amount of ferromagnetic material near to the regenerator and thereby alters the location and size of the magnetic field bump produced by the regenerator. Increasing the amount of ferromagnetic material causes the magnetic field bump to be moved inward (towards the plasma column – see, e.g., Figs. 19 to 21) to thereby primarily affect lower-energy particle orbits. Decreasing the amount of ferromagnetic material causes the magnetic field bump to be moved outward (away the plasma column) to thereby primarily affect higher-energy particle orbits.

[00118] The magnetic shims may be permanently screwed into the yoke and held in place there using screws or they may be controlled in real time. In this regard, movement or the magnetic shim(s) may be computer-controlled through a control system that is part of the particle therapy system. For example, the movement of each magnetic shim 730a, 730b may be controlled based on a rotational position of the particle accelerator, as measured by the rotation position of the gantry on which the particle accelerator is mounted. The various parameters used to set the magnetic shim location vis-à-vis the rotational position of the accelerator may be measured empirically, and programmed into the control system computer. One or more computer-controlled actuators may effect actual movement of the magnetic shim(s). Although only two magnetic shims are depicted, any number of magnetic shims may be used (e.g., one or more).

[00119] In some implementations, the magnetic shim(s) (e.g., the magnetic shim(s) described above) may instead be, or include, one or more miniature electromagnets, the current through which is controlled to affect the magnetic field produced by the regenerator in the manner described above. The current through the one or more elecrtomagnets may be computer-controlled through a control system that is part of the particle therapy system. For example, the current may be controlled based on a rotational position of the particle accelerator, as measured by the rotation position of the gantry on which the particle accelerator is mounted. The various parameters used to set the current vis-à-vis the rotational position of the accelerator may be measured empirically, and programmed into the control system computer.

[00120] In other implementations, adjusting the energy of particle beams reaching the extraction channel may be achieved by physically moving the cryostat to compensate for movement of the coil as a result of rotation. For example, the cryostat may be moved in a direction opposite to the direction that the coil moves. As above, the movement of the cryostat may be computer-controlled through a control system that is part of the particle therapy system. For example, the movement of cryostat may be controlled based on a rotational position of the particle accelerator, as measured by the rotation position of the gantry on which the particle accelerator is mounted. The various parameters used to set the movement of the cryostat vis-à-vis the rotational position of the gantry may be measured empirically, and programmed into the control system computer. One or more computer-controlled actuators may effect actual movement of the cryostat.

[00121] Referring to Fig. 24, for example, rotation of the accelerator may cause coils 709, 710 to move in the direction of arrow 735 within their respective chambers. In response, the position of cryostat 736 may be changed, e.g., cryostat 736 may be moved, e.g., in the direction of arrow 737 (e.g., in the opposite direction by an opposite amount). This movement causes a corresponding movement of coil 709, 710, thereby bringing coils 709, 710 back into their original position in proper alignment relative to the regenerator.

VARIABLE-ENERGY PARTICLE ACCELERATOR

[00122] The particle accelerator used in the example particle therapy systems described herein may be a variable-energy particle accelerator.

[00123] The energy of the extracted particle beam (the particle beam output from the accelerator) can affect the use of the particle beam during treatment. In some machines, the energy of the particle beam (or particles in the particle beam) does not increase after extraction. However, the energy may be reduced based on treatment needs after the extraction and before the treatment. Referring to Fig. 25, an example treatment system 910 includes an accelerator 912, e.g., a synchrocyclotron, from which a particle (e.g., proton) beam 914 having a variable energy is extracted to irradiate a target volume 924 of a body 922. Optionally, one or more additional devices, such as a scanning unit 916 or a scattering unit 916, one or more monitoring units 918, and an energy degrader 920, are placed along the irradiation direction 928. The devices intercept the cross-section of the extracted beam 914 and alter one or more properties of the extracted beam for the treatment.

[00124] A target volume to be irradiated (an irradiation target) by a particle beam for treatment typically has a three-dimensional configuration. In some examples, to carry-out the treatment, the target volume is divided into layers along the irradiation direction of the particle beam so that the irradiation can be done on a layer-by-layer basis. For certain types of particles, such as protons, the penetration depth (or which layer the beam reaches) within the target volume is largely determined by the energy of the particle beam. A particle beam of a given energy does not reach substantially beyond a corresponding penetration depth for that energy. To move the beam irradiation from one layer to another layer of the target volume, the energy of the particle beam is changed.

[00125] In the example shown in Fig. 25, the target volume 924 is divided into nine layers 926a-926i along the irradiation direction 928. In an example process, the irradiation starts from the deepest layer 926i, one layer at a time, gradually to the shallower layers and finishes with the shallowest layer 926a. Before application to the body 922, the energy of the particle beam 914 is controlled to be at a level to allow the particle beam to stop at a desired layer, e.g., the layer 926d, without substantially penetrating further into the body or the target volume, e.g., the layers 926e-926i or deeper into the body. In some examples, the desired energy of the particle beam 914 decreases as the treatment layer becomes shallower relative to the particle acceleration. In some examples, the beam energy difference for treating adjacent layers of the target volume 924 is about 3 MeV to about 100 MeV, e.g., about 10 MeV to about 80 MeV, although other differences may also be possible, depending on, e.g., the thickness of the layers and the properties of the beam.

[00126] The energy variation for treating different layers of the target volume 924 can be performed at the accelerator 912 (e.g., the accelerator can vary the energy) so that, in some implementations, no additional energy variation is required after the particle beam is extracted from the accelerator 912. So, the optional energy degrader 920 in the treatment system 10 may be eliminated from the system. In some implementations, the accelerator 912 can output particle beams having an energy that varies between about 100 MeV and about 300 MeV, e.g., between about

115 MeV and about 250 MeV. The variation can be continuous or non-continuous, e.g., one step at a time. In some implementations, the variation, continuous or non-continuous, can take place at a relatively high rate, e.g., up to about 50 MeV per second or up to about 20 MeV per second. Non-continuous variation can take place one step at a time with a step size of about 10 MeV to about 90 MeV.

[00127] When irradiation is complete in one layer, the accelerator 912 can vary the energy of the particle beam for irradiating a next layer, e.g., within several seconds or within less than one second. In some implementations, the treatment of the target volume 924 can be continued without substantial interruption or even without any interruption. In some situations, the step size of the non-continuous energy variation is selected to correspond to the energy difference needed for irradiating two adjacent layers of the target volume 924. For example, the step size can be the same as, or a fraction of, the energy difference.

[00128] In some implementations, the accelerator 912 and the degrader 920 collectively vary the energy of the beam 914. For example, the accelerator 912 provides a coarse adjustment and the degrader 920 provides a fine adjustment or vice versa. In this example, the accelerator 912 can output the particle beam that varies energy with a variation step of about 10-80 MeV, and the degrader 920 adjusts (e.g., reduces) the energy of the beam at a variation step of about 2-10 MeV.

[00129] The reduced use (or absence) of the energy degrader, which can include range shifters, helps to maintain properties and quality of the output beam from the accelerator, e.g., beam intensity. The control of the particle beam can be performed at the accelerator. Side effects, e.g., from neutrons generated when the particle beam passes the degrader 920 can be reduced or eliminated.

[00130] The energy of the particle beam 914 may be adjusted to treat another target volume 930 in another body or body part 922' after completing treatment in target volume 924. The target volumes 924, 930 may be in the same body (or patient), or may belong to different patients. It is possible that the depth D of the target volume 930 from a surface of body 922' is different from that of the target volume 924. Although some energy adjustment may be performed by the degrader 920, the degrader 912 may only reduce the beam energy and not increase the beam energy.

[00131] In this regard, in some cases, the beam energy required for treating target volume 930 is greater than the beam energy required to treat target volume 924. In such cases, the accelerator 912 may increase the output beam energy after treating the target volume 924 and before treating the target volume 930. In other cases, the beam energy required for treating target volume 930 is less than the beam energy required to treat target volume 924. Although the degrader 920 can reduce the energy, the accelerator 912 can be adjusted to output a lower beam energy to reduce or eliminate the use of the degrader 920. The division of the target volumes 924, 930 into layers can be different or the same. And the target volume 930 can be treated similarly on a layer by layer basis to the treatment of the target volume 924.

[00132] The treatment of the different target volumes 924, 930 on the same patient may be substantially continuous, e.g., with the stop time between the two volumes being no longer than about 30 minutes or less, e.g., 25 minutes or less, 20 minutes or less, 15 minutes or less, 10 minutes or less, 5 minutes or less, or 1 minute or less. As is explained herein, the accelerator 912 can be mounted on a

movable gantry and the movement of the gantry can move the accelerator to aim at different target volumes. In some situations, the accelerator 912 can complete the energy adjustment of the output beam 914 during the time the treatment system makes adjustment (such as moving the gantry) after completing the treatment of the target volume 924 and before starting treating the target volume 930. After the alignment of the accelerator and the target volume 930 is done, the treatment can begin with the adjusted, desired beam energy. Beam energy adjustment for different patients can also be completed relatively efficiently. In some examples, all adjustments, including increasing/reducing beam energy and/or moving the gantry are done within about 30 minutes, e.g., within about 25 minutes, within about 20 minutes, within about 15 minutes, within about 10 minutes or within about 5 minutes.

[00133] In the same layer of a target volume, an irradiation dose is applied by moving the beam across the two-dimensional surface of the layer (which is sometimes called scanning beam) using a scanning unit 916. Alternatively, the layer can be irradiated by passing the extracted beam through one or more scatterers of the scattering unit 16 (which is sometimes called scattering beam).

[00134] Beam properties, such as energy and intensity, can be selected before a treatment or can be adjusted during the treatment by controlling the accelerator 912 and/or other devices, such as the scanning unit/scatterer(s) 916, the degrader 920, and others not shown in the figures. In this example implementation, as in the example implementations described above, system 910 includes a controller 932, such as a computer, in communication with one or more devices in the system. Control can be based on results of the monitoring performed by the one or more monitors 918, e.g., monitoring of the beam intensity, dose, beam location in the target volume, etc. Although the monitors 918 are shown to be between the device 916 and the degrader 920, one or more monitors can be placed at other appropriate locations along the beam irradiation path. Controller 932 can also store a treatment plan for one or more target volumes (for the same patient and/or different patients). The treatment plan can be determined before the treatment starts and can include parameters, such as the shape of the target volume, the number of irradiation layers, the irradiation dose for each layer, the number of times each layer is irradiated, etc. The adjustment of a beam property within the system 910 can be performed based on the treatment plan. Additional adjustment can be made during the treatment, e.g., when deviation from the treatment plan is detected.

[00135] In some implementations, the accelerator 912 is configured to vary the energy of the output particle beam by varying the magnetic field in which the particle beam is accelerated. In an example implementation, one or more sets of coils receives variable electrical current to produce a variable magnetic field in the cavity. In some examples, one set of coils receives a fixed electrical current, while one or more other sets of coils receives a variable current so that the total current received by the coil sets varies. In some implementations, all sets of coils are superconducting. In other implementations, some sets of coils, such as the set for the fixed electrical current, are superconducting, while other sets of coils, such as the one or more sets for the variable current, are non-superconducting. In some examples, all sets of coils are non-superconducting.

[00136] Generally, the magnitude of the magnetic field is scalable with the magnitude of the electrical current. Adjusting the total electric current of the coils in a predetermined range can generate a magnetic field that varies in a corresponding, predetermined range. In some examples, a continuous adjustment of the electrical current can lead to a continuous variation of the magnetic field and a continuous variation of the output beam energy. Alternatively, when the electrical current applied to the coils is adjusted in a non-continuous, step-wise manner, the magnetic field and the output beam energy also varies accordingly in a non-continuous (step-wise) manner. The scaling of the magnetic field to the current can allow the variation of the beam energy to be carried out relatively precisely, although sometimes minor adjustment other than the input current may be performed.

[00137] In some implementations, to output particle beams having a variable energy, the accelerator 912 is configured to apply RF voltages that sweep over different ranges of frequencies, with each range corresponding to a different output beam energy. For example, if the accelerator 912 is configured to produce three different output beam energies, the RF voltage is capable of sweeping over three different ranges of frequencies. In another example, corresponding to continuous beam energy variations, the RF voltage sweeps over frequency ranges that continuously change. The different frequency ranges may have different lower frequency and/or upper frequency boundaries.

[00138] The extraction channel may be configured to accommodate the range of different energies produced by the variable-energy particle accelerator. Particle beams having different energies can be extracted from the accelerator 912 without altering the features of the regenerator that is used for extracting particle beams having a single energy. In other implementations, to accommodate the variable particle energy, the regenerator can be moved to disturb (e.g., change) different particle orbits in the manner described above and/or iron rods (magnetic shims) can be added or removed to change the magnetic field bump provided by the regenerator. More specifically, different particle energies will typically be at different particle orbits within the cavity. By moving the regenerator in the manner described herein, it is possible to intercept a particle orbit at a specified energy and thereby provide the correct perturbation of that orbit so that particles at the specified energy reach the extraction channel. In some implementations, movement of the regenerator (and/or addition/removal of magnetic shims) is performed in real-time to match real-time changes in the particle beam energy output by the accelerator. In other implementations, particle energy is adjusted on a per-treatment basis, and movement of the regenerator (and/or addition/removal of magnetic shims) is performed in advance of the treatment. In either case, movement of the regenerator (and/or addition/removal of magnetic shims) may be computer controlled. For example, a computer may control one or more motors that effect movement of the regenerator and/or magnetic shims.

[00139] In some implementations, the regenerator is implemented using one or more magnetic shims that are controllable to move to the appropriate location(s).

[00140] In some implementations, structure 715 (described above) is controlled to accommodate the different energies produced by the particle accelerator. For example, structure 715 may be rotated so that an appropriate thickness intercepts a particle beam having a particular energy. Structure 715 thus absorbs at least some of the energy in the particle beam, enabling the particle beam to traverse the extraction channel, as described above.

[00141] As an example, table 1 shows three example energy levels at which example accelerator 912 can output particle beams. The corresponding parameters for producing the three energy levels are also listed. In this regard, the magnet current refers to the total electrical current applied to the one or more coil sets in the accelerator 912; the maximum and minimum frequencies define the ranges in which the RF voltage sweeps; and "r" is the radial distance of a location to a center of the cavity in which the particles are accelerated.

Beam Energy (MeV)	Magnet Current (Amps)	Maximum Frequency (MHz)	Minimum Frequency (MHz)	Magnetic Field at r=0 mm (Tesla)	Magnetic Field at r=298 mm (Tesla)
250	1990	132	99	8.7	8.2
235	1920	128	97	8.4	8.0
211	1760	120	93	7.9	7.5

Table 1. Examples of beam energies and respective parameters.

[00142] Details that may be included in an example particle accelerator that produces charged particles having variable energies are described below. The accelerator can be a synchrocyclotron and the particles may be protons. The particles output as pulsed beams. The energy of the beam output from the particle accelerator can be varied during the treatment of one target volume in a patient, or between treatments of different target volumes of the same patient or different patients. In some implementations, settings of the accelerator are changed to vary the beam energy when no beam (or particles) is output from the accelerator. The energy variation can be continuous or non-continuous over a desired range.

[00143] Referring to the example shown in Fig. 1, the particle accelerator (synchrocyclotron 502), which may be a variable-energy particle accelerator like accelerator 912 described above, may be configured to particle beams that have a variable energy. The range of the variable energy can have an upper boundary that is about 200 MeV to about 300 MeV or higher, e.g., 200 MeV, about 205 MeV, about 210 MeV, about 215 MeV, about 220 MeV, about 225 MeV, about 230 MeV, about 235 MeV, about 240 MeV, about 245 MeV, about 250 MeV, about 255 MeV, about 260 MeV, about 265 MeV, about 270 MeV, about 275 MeV, about 280 MeV, about 285 MeV, about 290 MeV, about 295 MeV, or about 300 MeV or higher. The range can also have a lower boundary that is about 100 MeV or lower to about 200 MeV, e.g., about 100 MeV or lower, about 105 MeV, about 110 MeV, about 115 MeV, about 120 MeV, about 125 MeV, about 130 MeV, about 135 MeV, about 140 MeV, about 145 MeV, about 150 MeV, about 155 MeV, about 160 MeV, about 165 MeV, about 170 MeV, about 175 MeV, about 180 MeV, about 185 MeV, about 190 MeV, about 195 MeV, about 200 MeV.

[00144] In some examples, the variation is non-continuous and the variation step can have a size of about 10 MeV or lower, about 15 MeV, about 20 MeV, about 25 MeV, about 30 MeV, about 35 MeV, about 40 MeV, about 45 MeV, about 50 MeV, about 55 MeV, about 60 MeV, about 65 MeV, about 70 MeV, about 75 MeV, or about 80 MeV or higher. Varying the energy by one step size can take no more than 30 minutes, e.g., about 25 minutes or less, about 20 minutes or less, about 15 minutes or less, about 10 minutes or less, about 5 minutes or less, about 1 minute or less, or about 30 seconds or less. In other examples, the variation is continuous and the accelerator can adjust the energy of the particle beam at a relatively high rate, e.g., up to about 50 MeV per second, up to about 45 MeV per second, up to about 40 MeV per second, up to about 35 MeV per second, up to about 30 MeV per second, up to about 25 MeV per second, up to about 20 MeV per second, up to about 15 MeV per second, or up to about 10 MeV per second. The accelerator can be configured to adjust the particle energy both continuously and non-continuously. For example, a combination of the continuous and non-continuous variation can be used in a treatment of one target volume or in treatments of different target volumes. Flexible treatment planning and flexible treatment can be achieved.

[00145] A particle accelerator that outputs a particle beam having a variable energy can provide accuracy in irradiation treatment and reduce the number of additional devices (other than the accelerator) used for the treatment. For example, the use of degraders for changing the energy of an output particle beam may be reduced or eliminated. The properties of the particle beam, such as intensity, focus, etc. can be controlled at the particle accelerator and the particle beam can reach the target volume without substantial disturbance from the additional devices. The relatively high variation rate of the beam energy can reduce treatment time and allow for efficient use of the treatment system.

[00146] In some implementations, the accelerator, such as the synchrocyclotron 502 of Fig. 1, accelerates particles or particle beams to variable energy levels by varying the magnetic field in the accelerator, which can be achieved by varying the electrical current applied to coils for generating the magnetic field. As shown in Figs. 3, 4, 5, 6, and 7, example synchrocyclotron 10 (502 in Fig. 1) includes a magnet system that contains a particle source 90, a radiofrequency drive system 91, and a beam extraction system 38. Fig. 28 shows an example of a magnet system that may be used in a variable-energy accelerator. In this example implementation, the magnetic field established by the magnet system 1012 can vary by about 5% to about 35 % of a maximum value of the magnetic field that two sets of coils 40a and 40b, and 42a and 42b are capable of generating. The magnetic field established by the magnet system has a shape appropriate to maintain focus of a contained proton beam using a combination of the two sets of coils and a pair of shaped ferromagnetic (e.g., low carbon steel) structures, examples of which are provided above.

[00147] Each set of coils may be a split pair of annular coils to receive electrical current. In some situations, both sets of coils are superconducting. In other situations, only one set of the coils is superconducting and the other set is non-superconducting or normal conducting (also discussed further below). It is also possible that both sets of coils are non-superconducting. Suitable superconducting materials for use in the coils include niobium-3 tin (Nb3Sn) and/or niobium-titanium. Other normal conducting materials can include copper. Examples of the coil set constructions are described further below.

[00148] The two sets of coils can be electrically connected serially or in parallel. In some implementations, the total electrical current received by the two sets of coils can include about 2 million ampere turns to about 10 million ampere turns, e.g., about 2.5 to about 7.5 million ampere turns or about 3.75 million ampere turns to about 5 million ampere turns. In some examples, one set of coils is configured to receive a fixed (or constant) portion of the total variable electrical current, while the other set of coils is configured to receive a variable portion of the total electrical current. The total electrical current of the two coil sets varies with the variation of the current in one coil set. In other situations, the electrical current applied to both sets of coils can vary. The variable total current in the two sets of coils can generate a magnetic field having a variable magnitude, which in turn varies the acceleration pathways of the particles and produces particles having variable energies.

[00149] Generally, the magnitude of the magnetic field generated by the coil(s) is scalable to the magnitude of the total electrical current applied to the coil(s). Based on the scalability, in some implementations, linear variation of the magnetic field strength can be achieved by linearly changing the total current of the coil sets. The total current can be adjusted at a relatively high rate that leads to a relatively high-rate adjustment of the magnetic field and the beam energy.

[00150] In the example reflected in Table 1 above, the ratio between values of the current and the magnetic field at the geometric center of the coil rings is: 1990:8.7 (approximately 228.7:1); 1920:8.4 (approximately 228.6:1); 1760:7.9 (approximately 222.8:1). Accordingly, adjusting the magnitude of the total current applied to a superconducting coil(s) can proportionally (based on the ratio) adjust the magnitude of the magnetic field.

[00151] The scalability of the magnetic field to the total electrical current in the example of Table 1 is also shown in the plot of Fig. 26, where BZ is the magnetic field along the Z direction; and R is the radial distance measured from a geometric center of the coil rings along a direction perpendicular to the Z direction. The magnetic field has the highest value at the geometric center, and decreases as the distance R increases. The curves 1035, 1037 represent the magnetic field generated by the same coil sets receiving different total electrical current: 1760 Amperes and 1990 Amperes, respectively. The corresponding energies of the extracted particles are 211 MeV and 250 MeV, respectively. The two curves 1035, 1037 have substantially the same shape and the different parts of the curves 1035, 1037 are substantially parallel. As a result, either the curve 1035 or the curve 1037 can be linearly shifted to substantially match the other curve, indicating that the magnetic field is scalable to the total electrical current applied to the coil sets.

[00152] In some implementations, the scalability of the magnetic field to the total electrical current may not be perfect. For example, the ratio between the magnetic field and the current calculated based on the example shown in table 1 is not constant. Also, as shown in Fig. 26, the linear shift of one curve may not perfectly match the other curve. In some implementations, the total current is applied to the coil sets under the assumption of perfect scalability. The target magnetic field (under the assumption of perfect scalability) can be generated by additionally altering the features, e.g., geometry, of the coils to counteract the imperfection in the scalability. As one example, ferromagnetic (e.g., iron) rods (magnetic shims) can be inserted or removed from one or both of the magnetic structures. The features of the coils can be altered at a relatively high rate so that the rate of the magnetic field adjustment is not substantially affected as compared to the situation in which the scalability is perfect and only the electrical current needs to be adjusted. In the example of iron rods, the rods can be added or removed at the time scale of seconds or minutes, e.g., within 5 minutes, within 1 minute, less than 30 seconds, or less than 1 second.

[00153] In some implementations, settings of the accelerator, such as the current applied to the coil sets, can be chosen based on the substantial scalability of the magnetic field to the total electrical current in the coil sets.

[00154] Generally, to produce the total current that varies within a desired range, any combination of current applied to the two coil sets can be used. In an example, the coil set 42a, 42b can be configured to receive a fixed electrical current corresponding to a lower boundary of a desired range of the magnetic field. In the example shown in table 1, the fixed electrical current is 1760 Amperes. In addition, the coil set 40a, 40b can be configured to receive a variable electrical current having an upper boundary corresponding to a difference between an upper boundary and a lower boundary of the desired range of the magnetic field. In the example shown in table 1, the coil set 40a, 40b is configured to receive electrical current that varies between 0 Ampere and 230 Amperes.

[00155] In another example, the coil set 42a, 42b can be configured to receive a fixed electrical current corresponding to an upper boundary of a desired range of the magnetic field. In the example shown in table 1, the fixed current is 1990 Amperes. In addition, the coil set 40a, 40b can be configured to receive a variable electrical current having an upper boundary corresponding to a difference between a lower boundary and an upper boundary of the desired range of the magnetic field. In the example shown in table 1, the coil set 40a, 40b is configured to receive electrical current that varies between -230 Ampere and 0 Ampere.

[00156] The total variable magnetic field generated by the variable total current for accelerating the particles can have a maximum magnitude greater than 4 Tesla, e.g., greater than 5 Tesla, greater than 6 Tesla, greater than 7 Tesla, greater than 8 Tesla, greater than 9 Tesla, or greater than 10 Tesla, and up to about 20 Tesla or higher, e.g., up to about 18 Tesla, up to about 15 Tesla, or up to about 12 Tesla. In some implementations, variation of the total current in the coil sets can vary the magnetic field by about 0.2 Tesla to about 4.2 Tesla or more, e.g., about 0.2 Tesla to about 1.4 Tesla or about 0.6 Tesla to about 4.2 Tesla. In some situations, the amount of variation of the magnetic field can be proportional to the maximum magnitude.

[00157] Fig. 27 shows an example RF structure for sweeping the voltage on the dee plate 100 over an RF frequency range for each energy level of the particle beam, and for varying the frequency range when the particle beam energy is varied. The semicircular surfaces 103, 105 of the dee plate 100 are connected to an inner conductor 1300 and housed in an outer conductor 1302. The high voltage is applied to the dee plate 100 from a power source (not shown, e.g., an oscillating voltage input) through a power coupling device 1304 that couples the power source to the inner conductor. In some implementations, the coupling device 1304 is positioned on the inner conductor 1300 to provide power transfer from the power source to the dee plate 100. In addition, the dee plate 100 is coupled to variable reactive elements 1306, 1308 to perform the RF frequency sweep for each particle energy level, and to change the RF frequency range for different particle energy levels.

[00158] The variable reactive element 1306 can be a rotating capacitor that has multiple blades 1310 rotatable by a motor (not shown). By meshing or unmeshing the blades 1310 during each cycle of RF sweeping, the capacitance of the RF structure changes, which in turn changes the resonant frequency of the RF structure. In some implementations, during each quarter cycle of the motor, the blades 1310 mesh with the each other. The capacitance of the RF structure increases and the resonant frequency decreases. The process reverses as the blades 1310 unmesh. As a result, the power required to generate the high voltage applied to the dee plate 103 and necessary to accelerate the beam can be reduced by a large factor. In some implementations, the shape of the blades 1310 is machined to form the required dependence of resonant frequency on time.

[00159] The RF frequency generation is synchronized with the blade rotation by sensing the phase of the RF voltage in the resonator, keeping the alternating voltage on the dee plates close to the resonant frequency of the RF cavity. (The dummy dee is grounded and is not shown in Fig. 27).

[00160] The variable reactive element 1308 can be a capacitor formed by a plate 1312 and a surface 1316 of the inner conductor 1300. The plate 1312 is movable along a direction 1314 towards or away from the surface 1316. The capacitance of the capacitor changes as the distance D between the plate 1312 and the surface 1316 changes. For each frequency range to be swept for one particle energy, the distance D is at a set value, and to change the frequency range, the plate 1312 is moved corresponding to the change in the energy of the output beam.

[00161] In some implementations, the inner and outer conductors 1300, 1302 are formed of a metallic material, such as copper, aluminum, or silver. The blades 1310 and the plate 1312 can also be formed of the same or different metallic materials as the conductors 1300, 1302. The coupling device 1304 can be an electrical conductor. The variable reactive elements 1306, 1308 can have other forms and can couple to the dee plate 100 in other ways to perform the RF frequency sweep and the frequency range alteration. In some implementations, a single variable reactive element can be configured to perform the functions of both the variable reactive elements 1306, 1308. In other implementations, more than two variable reactive elements can be used.

[00162] Any of the features described herein may be configured for use with a variable-energy particle accelerator, such as that described above.

[00163] Any two more of the foregoing implementations may be used in an appropriate combination to affect the energy of a particle beam in the extraction channel. Likewise, individual features of any two more of the foregoing implementations may be used in an appropriate combination for the same purpose.

[00164] Elements of different implementations described herein may be combined to form other implementations not specifically set forth above. Elements may be left out of the processes, systems, apparatus, etc., described herein without adversely affecting their operation. Various separate elements may be combined into one or more individual elements to perform the functions described herein.

[00165] The example implementations described herein are not limited to use with a particle therapy system or to use with the example particle therapy systems described herein. Rather, the example implementations can be used in any appropriate system that directs accelerated particles to an output.

[00166] Additional information concerning the design of an example implementation of a particle accelerator that may be used in a system as described herein can be found in U.S. Provisional Application No. 60/760,788, entitled "High-Field Superconducting Synchrocyclotron" and filed January 20, 2006; U.S. Patent Application No. 11/463,402, entitled "Magnet Structure For Particle Acceleration" and filed August 9, 2006; and U.S. Provisional Application No. 60/850,565, entitled "Cryogenic Vacuum Break Pneumatic Thermal Coupler" and filed October 10, 2006, all of which are incorporated herein by reference.

[00167] The following applications are incorporated by reference into the subject application: the U.S. Provisional Application entitled "CONTROLLING INTENSITY OF A PARTICLE BEAM" (Application No. 61/707,466), the U.S. Provisional Application entitled "ADJUSTING ENERGY OF A PARTICLE BEAM" (Application No. 61/707,515), the U.S. Provisional Application entitled "ADJUSTING COIL POSITION" (Application No. 61/707,548), the U.S. Provisional Application entitled "FOCUSING A PARTICLE BEAM USING MAGNETIC FIELD FLUTTER" (Application No. 61/707,572), the U.S. Provisional Application entitled "MAGNETIC FIELD REGENERATOR" (Application No. 61/707,590), the U.S. Provisional Application entitled "FOCUSING A PARTICLE BEAM" (Application No. 61/707,704), the U.S. Provisional Application entitled "CONTROLLING PARTICLE THERAPY (Application No. 61/707,624), and the U.S. Provisional Application entitled "CONTROL SYSTEM FOR A PARTICLE ACCELERATOR" (Application No. 61/707,645).

[00168] The following are also incorporated by reference into the subject application: U.S. Patent No. 7,728,311 which issued on June 1, 2010, U.S. Patent Application No. 11/948,359 which was filed on November 30, 2007, U.S. Patent Application No. 12/275,103 which was filed on November 20, 2008, U.S. Patent Application No. 11/948,662 which was filed on November 30, 2007, U.S. Provisional Application No. 60/991,454 which was filed on November 30, 2007, U.S. Patent No. 8,003,964 which issued on August 23, 2011, U.S. Patent No. 7,208,748 which issued on April 24, 2007, U.S. Patent No. 7,402,963 which issued on July 22, 2008, U.S. Patent Application No. 13/148,000 filed February 9, 2010, U.S. Patent Application No. 11/937,573 filed on November 9, 2007, U.S. Patent Application No.

11/187,633, titled "A Programmable Radio Frequency Waveform Generator for a Synchrocyclotron," filed July 21, 2005, U.S. Provisional Application No. 60/590,089, filed on July 21, 2004, U.S. Patent Application No. 10/949,734, titled "A Programmable Particle Scatterer for Radiation Therapy Beam Formation", filed September 24, 2004, and U.S. Provisional Application No.60/590,088, filed July 21, 2005.

[00169] Any features of the subject application may be combined with one or more appropriate features of the following: the U.S. Provisional Application entitled "CONTROLLING INTENSITY OF A PARTICLE BEAM" (Application No. 61/707,466), the U.S. Provisional Application entitled "ADJUSTING ENERGY OF A PARTICLE BEAM" (Application No. 61/707,515), the U.S. Provisional Application entitled "ADJUSTING COIL POSITION" (Application No. 61/707,548), the U.S. Provisional Application entitled "FOCUSING A PARTICLE BEAM USING MAGNETIC FIELD FLUTTER" (Application No. 61/707,572), the U.S. Provisional Application entitled "MAGNETIC FIELD REGENERATOR" (Application No. 61/707,590), the U.S. Provisional Application entitled "FOCUSING A PARTICLE BEAM" (Application No. 61/707,704), the U.S. Provisional Application entitled "CONTROLLING PARTICLE THERAPY (Application No. 61/707,624), and the U.S. Provisional Application entitled "CONTROL SYSTEM FOR A PARTICLE ACCELERATOR" (Application No. 61/707,645), U.S. Patent No. 7,728,311 which issued on June 1, 2010, U.S. Patent Application No. 11/948,359 which was filed on November 30, 2007, U.S. Patent Application No. 12/275,103 which was filed on November 20, 2008, U.S. Patent Application No. 11/948,662 which was filed on November 30, 2007, U.S. Provisional Application No. 60/991,454 which was filed on

November 30, 2007, U.S. Patent Application No. 13/907,601, which was filed on May 31, 2013, U.S. Patent Application No. 13/916,401, filed on June 12, 2013, U.S. Patent No. 8,003,964 which issued on August 23, 2011, U.S. Patent No. 7,208,748 which issued on April 24, 2007, U.S. Patent No. 7,402,963 which issued on July 22, 2008, U.S. Patent Application No. 13/148,000 filed February 9, 2010, U.S. Patent Application No. 11/937,573 filed on November 9, 2007, U.S. Patent Application No. 11/187,633, titled "A Programmable Radio Frequency Waveform Generator for a Synchrocyclotron," filed July 21, 2005, U.S. Provisional Application No. 60/590,089, filed on July 21, 2004, U.S. Patent Application No. 10/949,734, titled "A Programmable Particle Scatterer for Radiation Therapy Beam Formation", filed September 24, 2004, and U.S. Provisional Application No. 60/590,088, filed July 21, 2005.

[00170] Except for the provisional application from which this patent application claims priority and the documents incorporated by reference above, no other documents are incorporated by reference into this patent application.

[00171] Other implementations not specifically described herein are also within the scope of the following claims.

[00172] What is claimed is:

1. A particle accelerator comprising:

a coil to provide a magnetic field to a cavity;

a particle source to provide a plasma column to the cavity;

a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, the magnetic field causing particles accelerated from the plasma column to move orbitally within the cavity;

an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and

a structure arranged proximate to the extraction channel to change an energy level of the received particles.

2. The particle accelerator of claim 1, wherein the structure has multiple thicknesses; and

wherein the structure is movable relative to the extraction channel to place one of the multiple thicknesses in a path of the received particles.

3. The particle accelerator of claim 2, wherein the structure is wheel-shaped and is rotatable within the extraction channel.

4. The particle accelerator of claim 2, wherein the structure has variable thickness ranging from a maximum thickness to a minimum thickness.

5. The particle accelerator of claim 1, wherein the particle accelerator is rotatable relative to a fixed position; and

wherein the particle accelerator further comprises a control system to control movement of the structure based on a rotational position of the particle accelerator.

6. The particle accelerator of claim 1, further comprising:

a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel.

7. The particle accelerator of claim 1, wherein the structure comprises at least one of the following materials: beryllium, carbon and plastic.

8. A proton therapy system comprising:

the particle accelerator of claim 1, wherein the particles comprise protons; and

a gantry on which the particle accelerator is mounted, the gantry being rotatable relative to a patient position;

wherein protons are output essentially directly from the particle accelerator to the patient position.

9. A particle accelerator comprising:

a coil to provide a magnetic field to a cavity;

a particle source to provide a plasma column to the cavity;

a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, the magnetic field causing particles accelerated from the plasma column to move orbitally within the cavity;

an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and

a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel, the regenerator being movable within the cavity relative to orbits of the particles.

10. The particle accelerator of claim 9, wherein the regenerator is configured to move radially relative to an approximate center of the cavity.

11. The particle accelerator of claim 10, further comprising:

an actuator to move the regenerator in response to a control signal.

12. The particle accelerator of claim 11, wherein the particle accelerator is rotatable relative to a fixed position; and

wherein the particle accelerator further comprises a control system to generate the control signal to control movement of the regenerator based on a rotational position of the particle accelerator.

13. The particle accelerator of claim 9, wherein the regenerator comprises a ferromagnetic material.

14. A proton therapy system comprising:

the particle accelerator of claim 9, wherein the particles comprise protons; and

a gantry on which the particle accelerator is mounted, the gantry being rotatable relative to a patient position;

wherein protons are output essentially directly from the particle accelerator to the patient position.

15. A particle accelerator comprising:

a coil to provide a magnetic field to a cavity;

a particle source to provide a plasma column to the cavity;

a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, the magnetic field causing particles accelerated from the plasma column to move orbitally within the cavity;

an enclosure containing an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and

a regenerator to adjust the magnetic field within the cavity to thereby change successive orbits of the particles accelerated from the plasma column so that, eventually, the particles output to the extraction channel;

wherein the enclosure comprise magnetic structures, at least one of the magnetic structures having a slot therein, the slot containing a magnetic shim that is ferromagnetic and movable within the slot, the magnetic shim being movable relative to the regenerator to affect an amount by which the regenerator adjusts the magnetic field.

16. The particle accelerator of claim 15, wherein the at least one of the magnetic structures has multiple slots therein, each slot containing a magnetic shim that is ferromagnetic and movable within the slot, each magnetic shim being movable relative to the regenerator to affect an amount by which the regenerator adjusts the magnetic field.

17. The particle accelerator of claim 15, wherein the particle accelerator is rotatable relative to a fixed position; and

wherein the particle accelerator further comprises a control system to generate a control signal to control movement of the magnetic shim based on a rotational position of the particle accelerator.

18. The particle accelerator of claim 15, wherein the magnetic shim comprises an electromagnet.

19. A proton therapy system comprising:

the particle accelerator of claim 15, wherein the particles comprise protons; and a gantry on which the particle accelerator is mounted, the gantry being rotatable relative to a patient position;

wherein protons are output essentially directly from the particle accelerator to the patient position.

20. A particle accelerator comprising:

a cryostat comprising a superconducting coil, the superconducting coil conducting a current that generates a magnetic field;

magnetic structures adjacent to the cryostat, the cryostat being attached to the magnetic structures, the magnetic structures containing a cavity;

a particle source to provide a plasma column to the cavity;

a voltage source to provide a radio frequency (RF) voltage to the cavity to accelerate particles from the plasma column, the magnetic field causing particles accelerated from the plasma column to move orbitally within the cavity;

an extraction channel to receive the particles accelerated from the plasma column and to output the received particles from the cavity; and

an actuator that is controllable to move the cryostat relative to the magnetic structures.

21. The particle accelerator of claim 20, wherein the particle accelerator is rotatable relative to a fixed position; and

wherein the particle accelerator further comprises a control system to generate a control signal to control the actuator based on a rotational position of the particle accelerator.

22. The particle accelerator of claim 21, wherein the actuator is controlled to control movement of the cryostat so as to compensate for effects of gravity on the superconducting coil.

23. A proton therapy system comprising:

the particle accelerator of claim 20, wherein the particles comprise protons; and

a gantry on which the particle accelerator is mounted, the gantry being rotatable relative to a patient position;

wherein protons are output essentially directly from the particle accelerator to the patient position.

24. A variable-energy particle accelerator comprising:

magnetic structures defining a cavity in which particles are accelerated for output as a particle beam that has a selected energy from among a range of energies;

an extraction channel to receive the particle beam; and

a structure proximate to the extraction channel to intercept the particle beam prior to the particle beam entering the extraction channel, the structure being movable based on the selected energy, the structure to absorb at least some energy of the particle beam prior to the particle beam entering the extraction channel.

25. The variable-energy particle accelerator of claim 24, wherein the structure comprises a wheel having varying thickness, where different thicknesses are capable of absorbing different amounts of energy.

26. The variable-energy particle accelerator of claim 24, further comprising:

a magnetic regenerator to implement a magnetic field bump at a particle orbit that corresponds to the selected energy;

wherein the magnetic regenerator is movable based on movement of the variable-energy particle accelerator.

27. The variable-energy particle accelerator of claim 24, further comprising:

a magnetic regenerator to implement a magnetic field bump at a particle orbit that corresponds to the selected energy;

wherein the magnetic regenerator is movable to intercept a particle orbit having the selected energy.

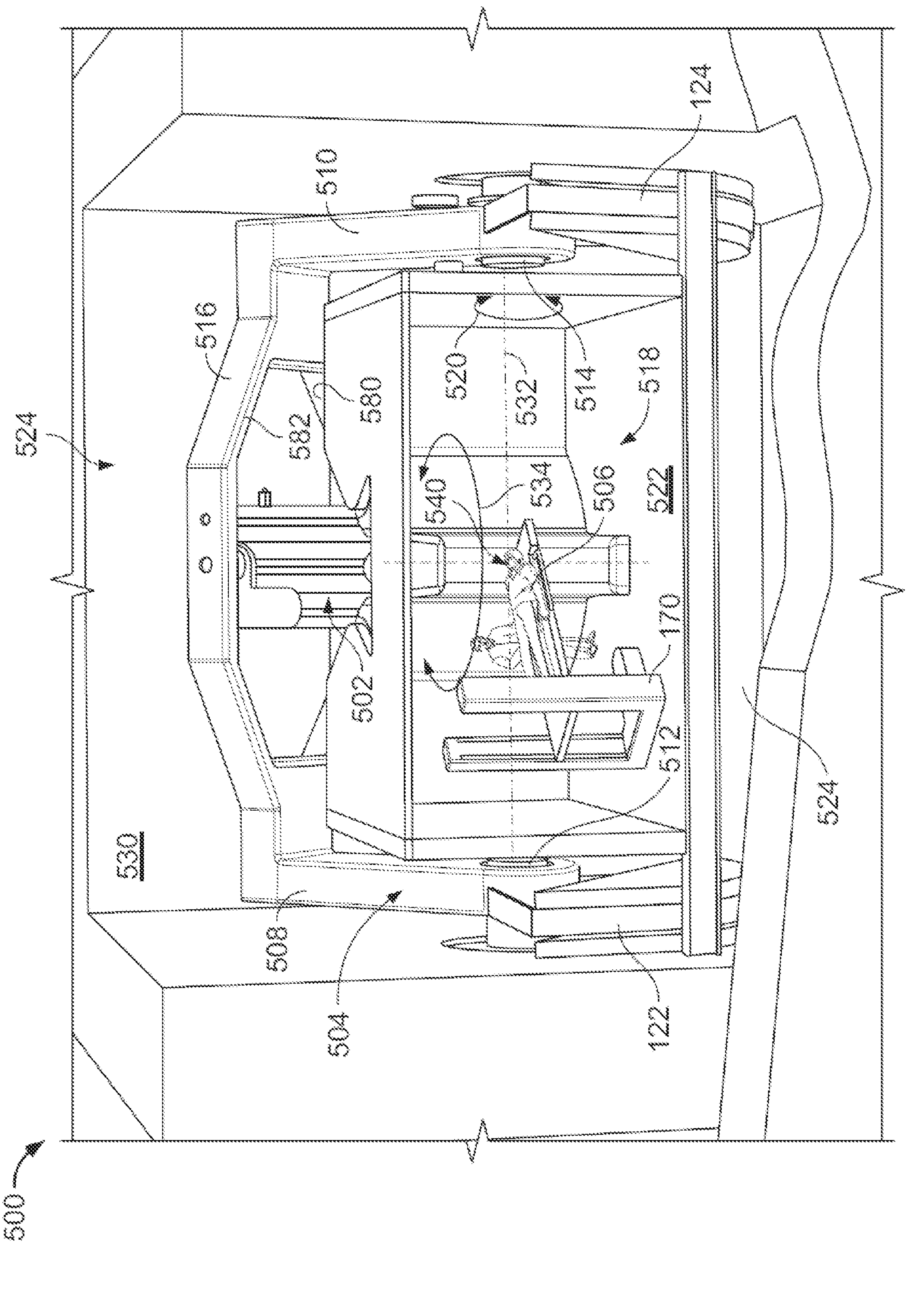
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FIG. 1

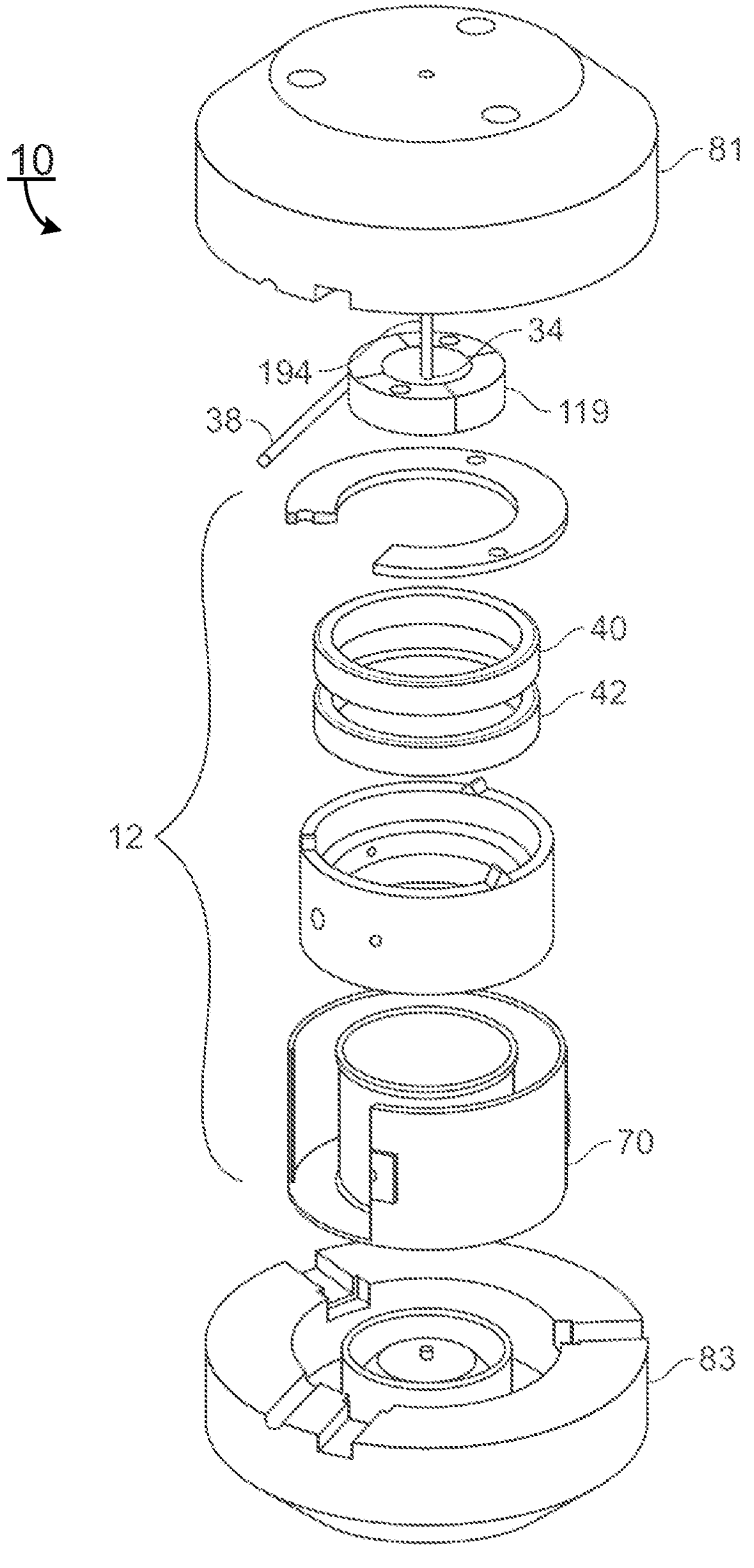

FIG. 2

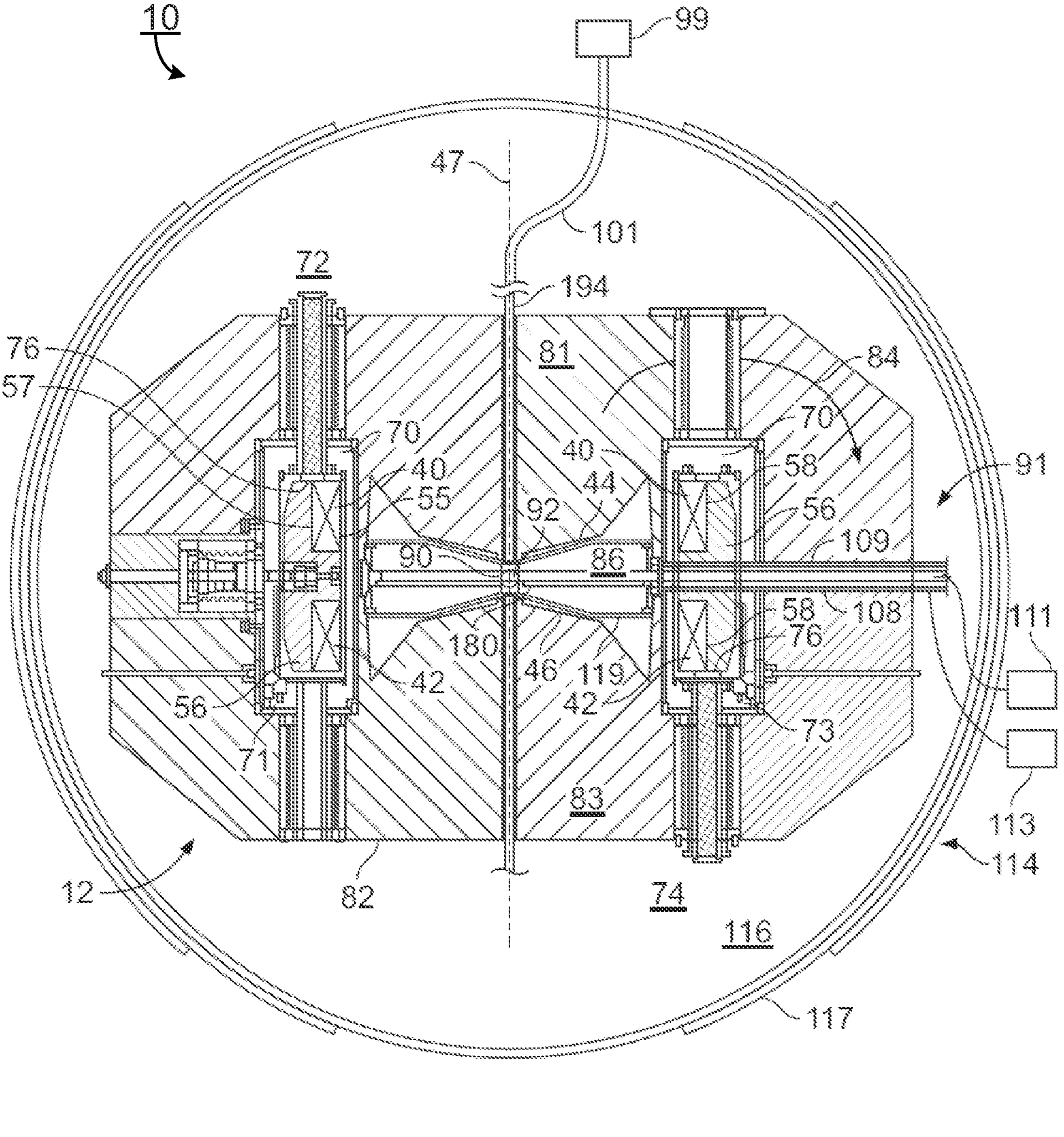
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FIG. 3

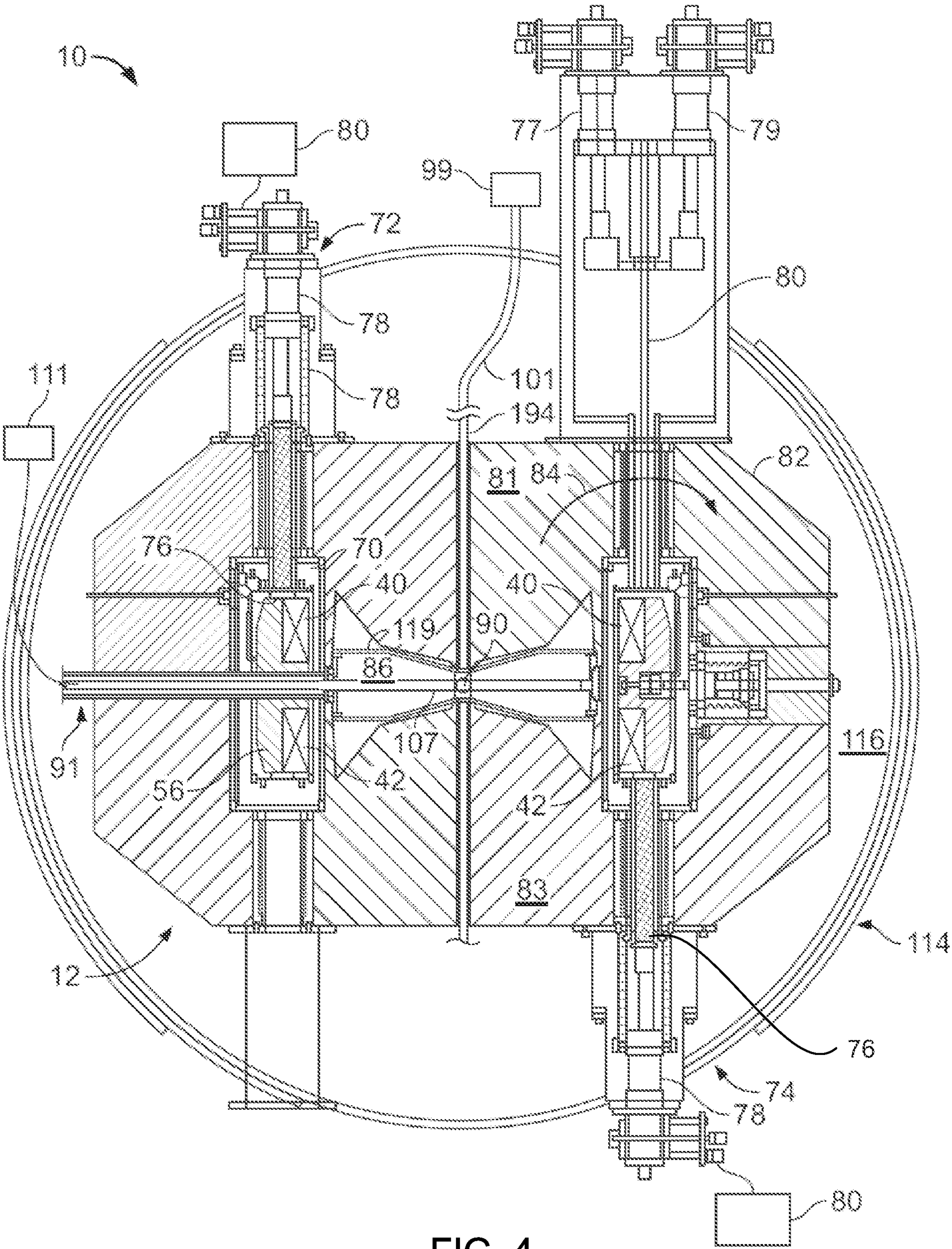
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FIG. 4

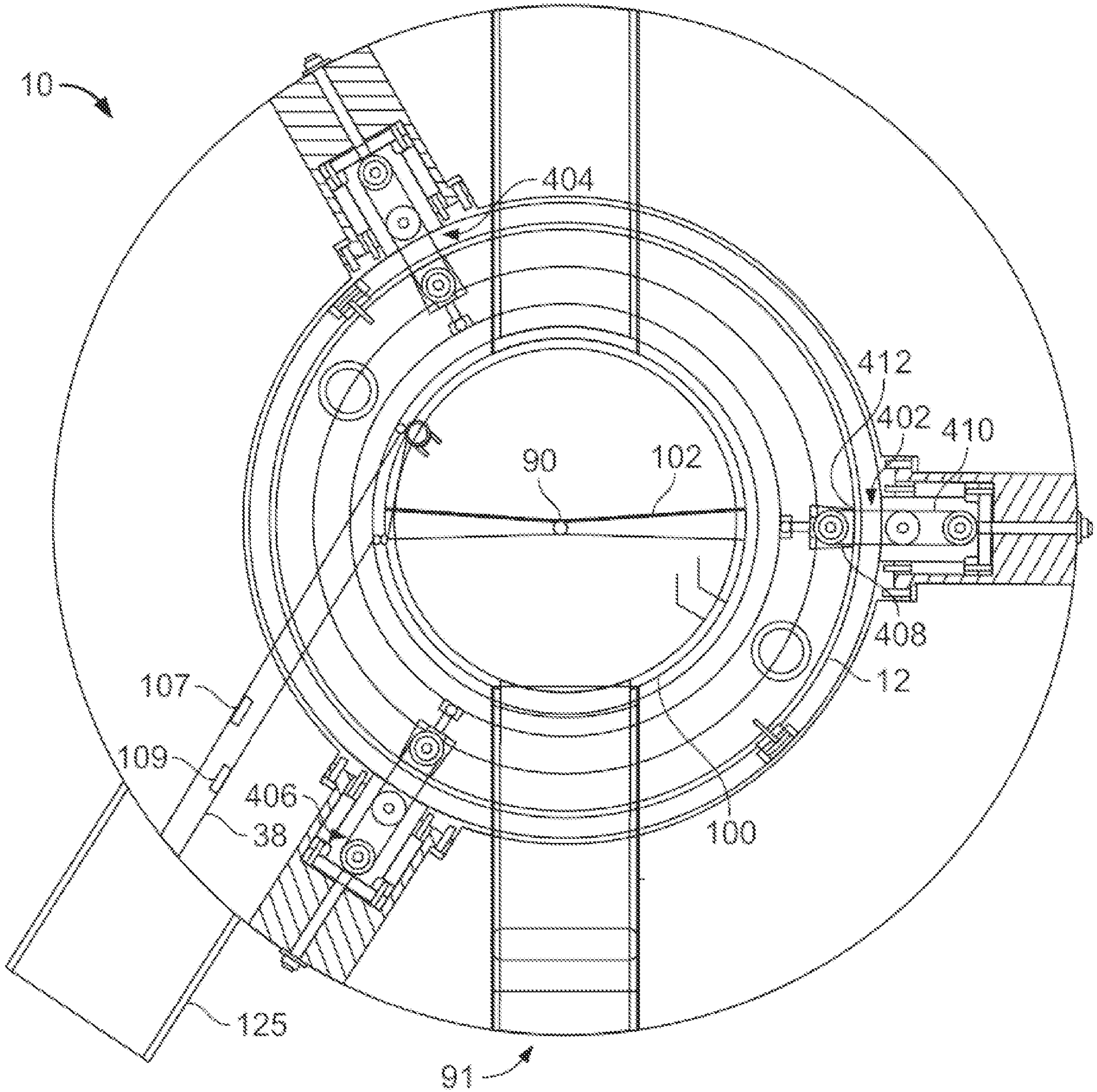

FIG. 5

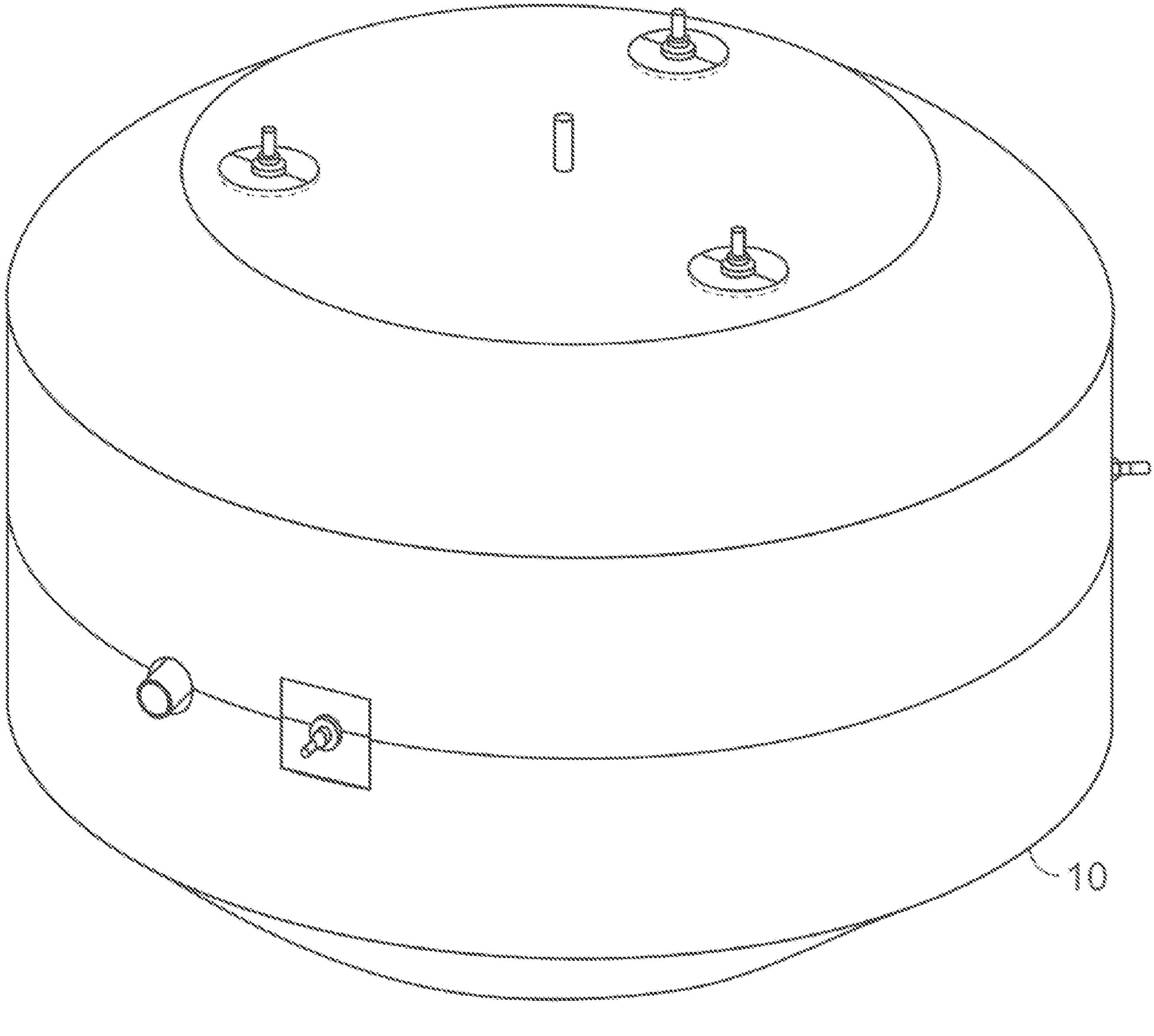

FIG. 6

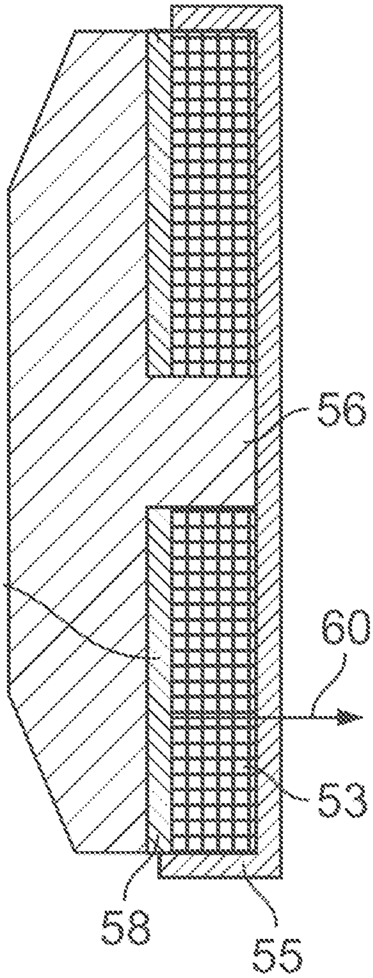

FIG. 7

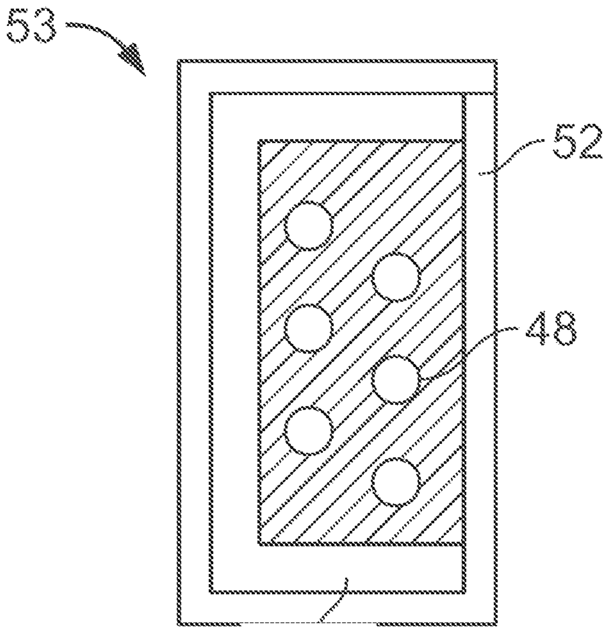

FIG. 8

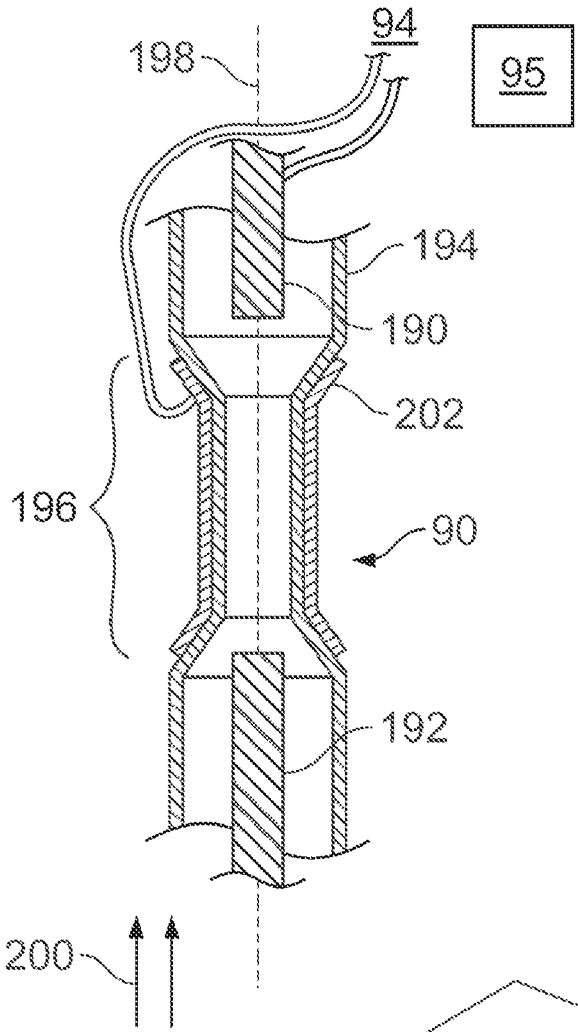

FIG. 9

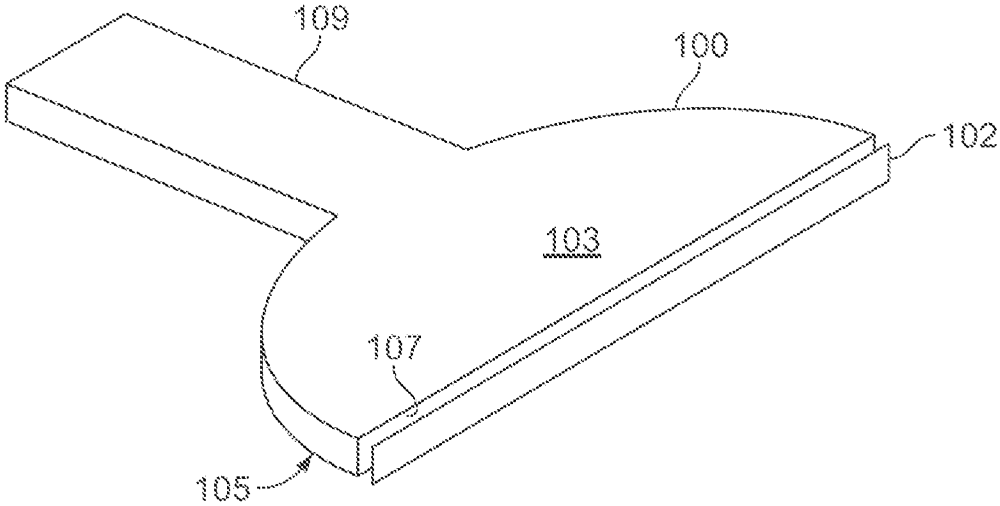

FIG. 10

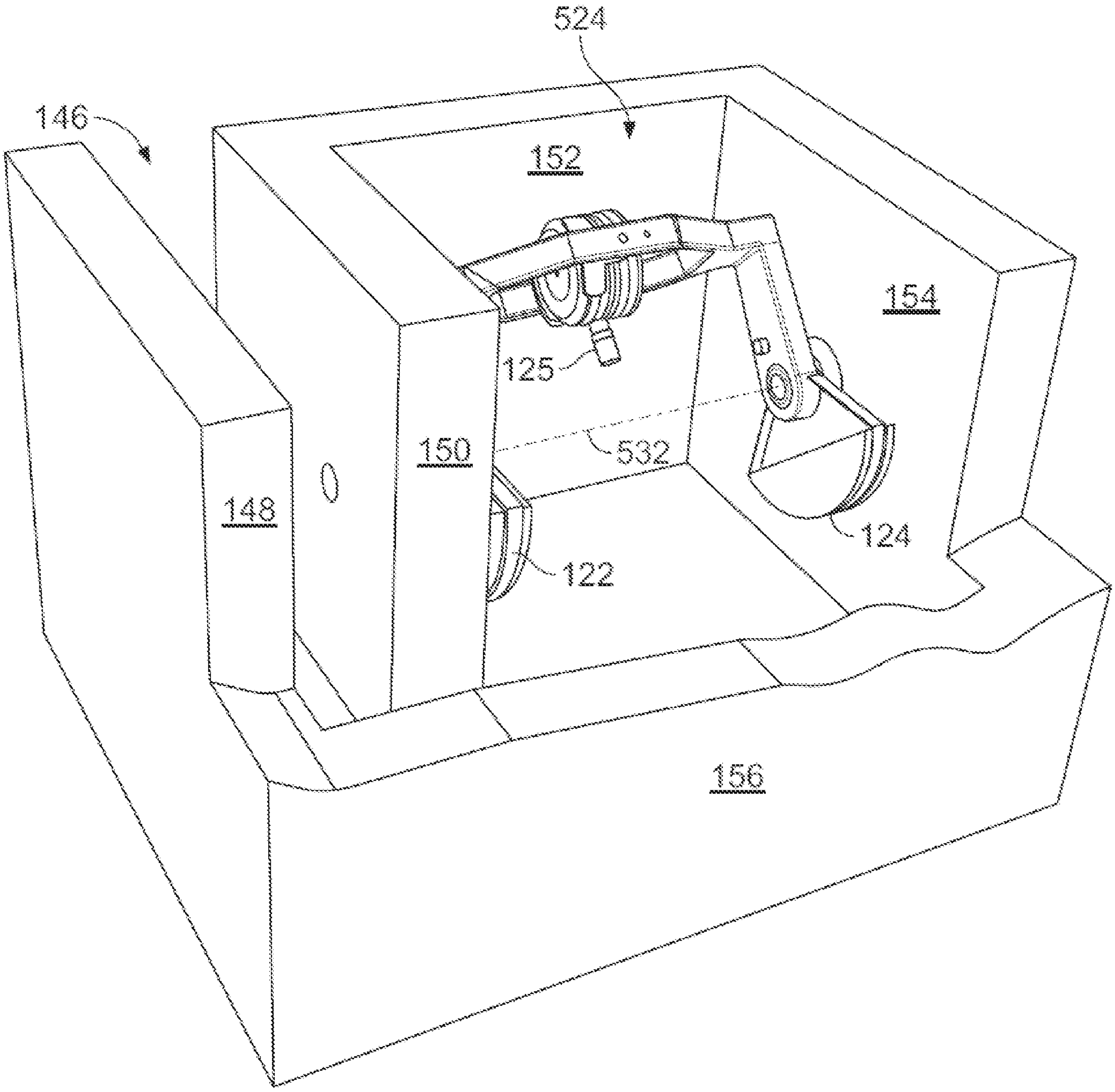

FIG. 11

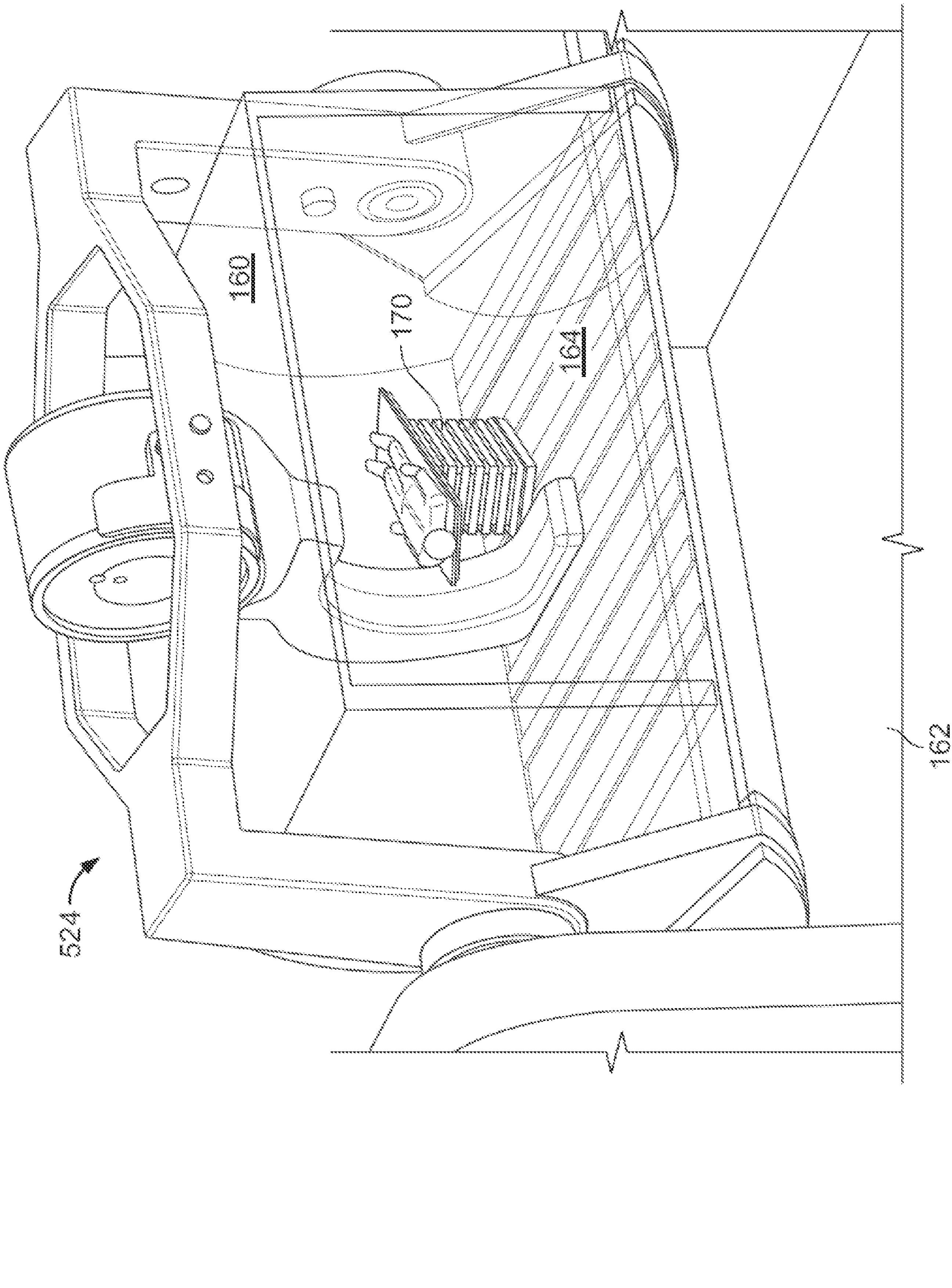

FIG. 12

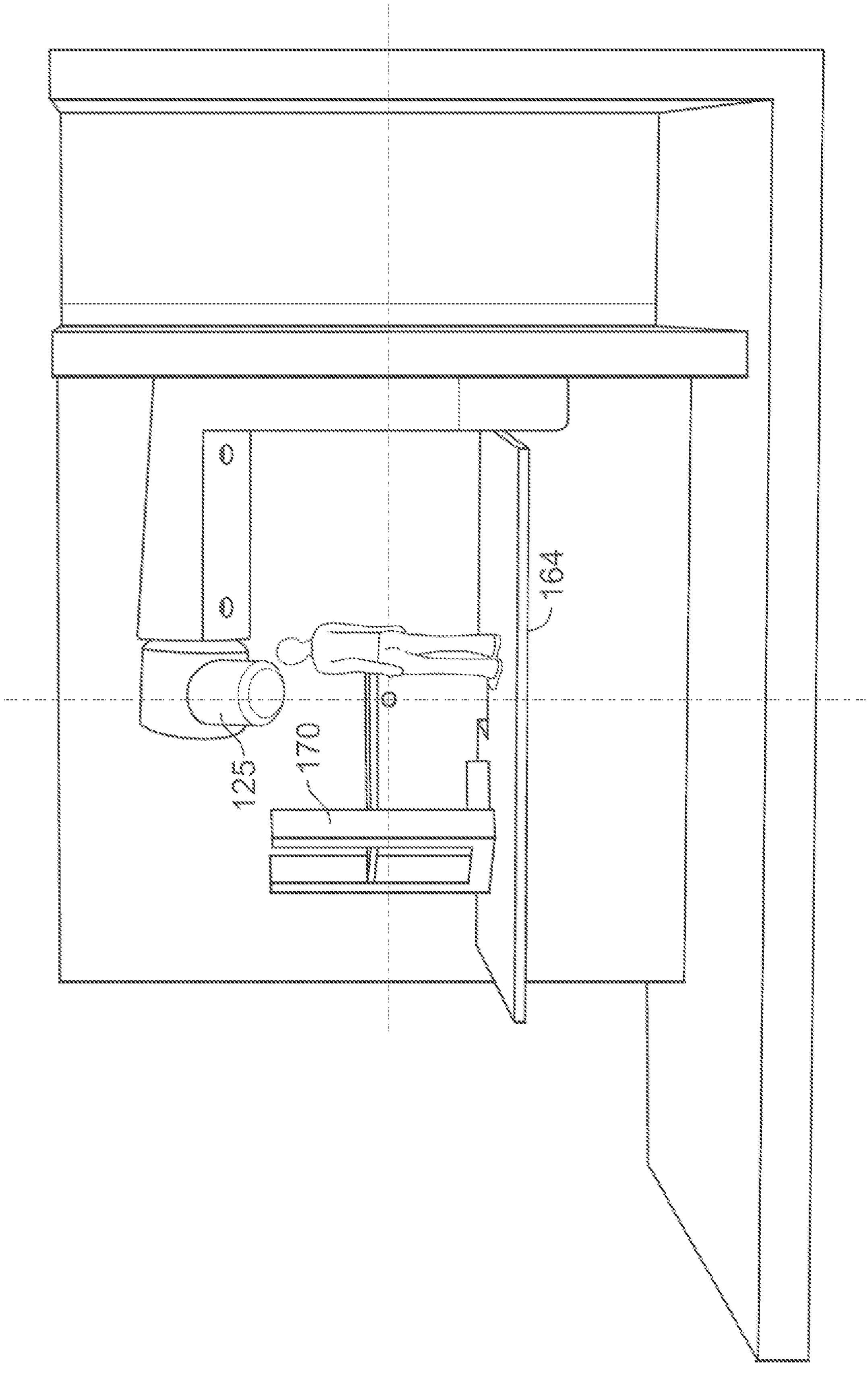

FIG. 13

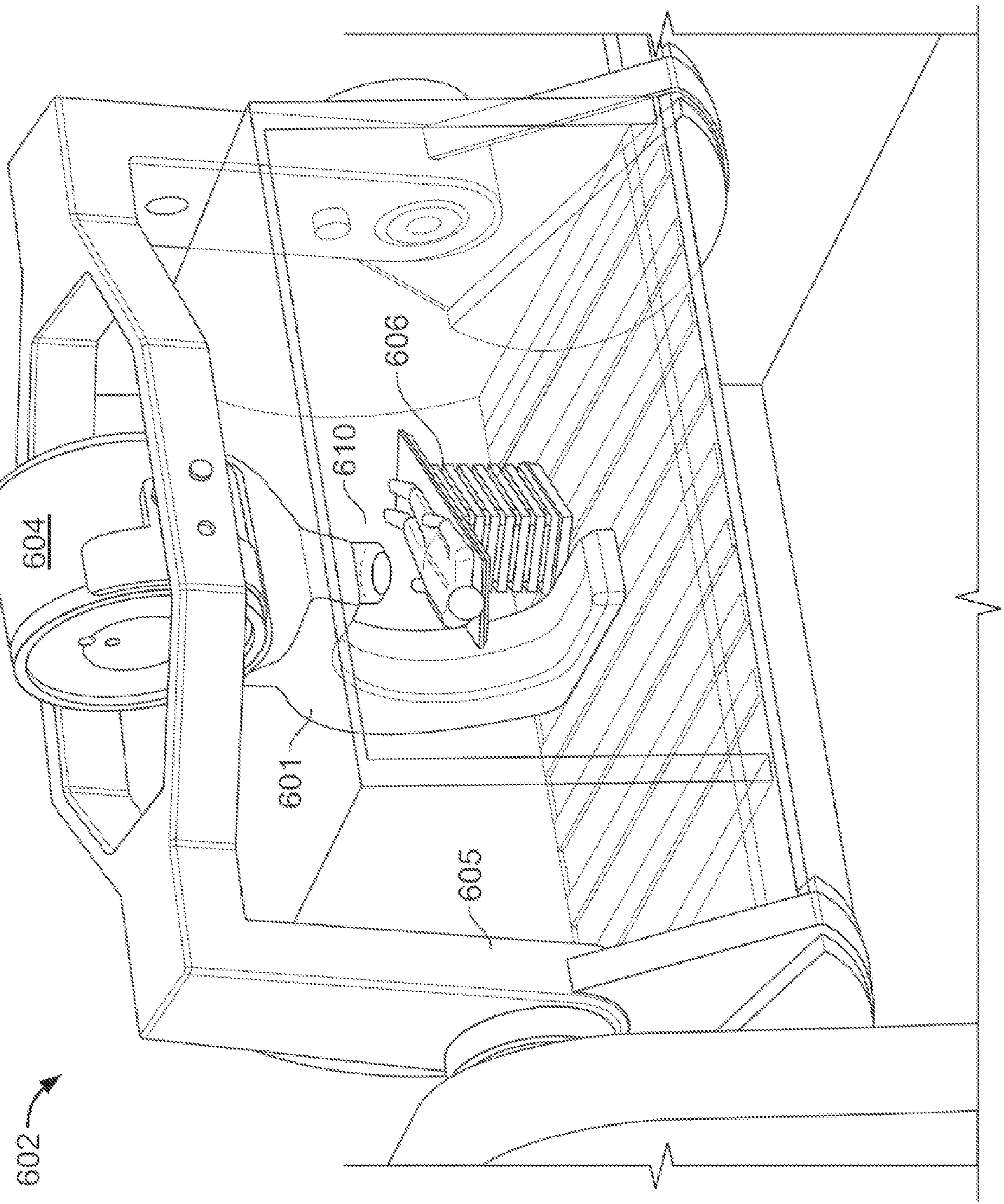

FIG. 14

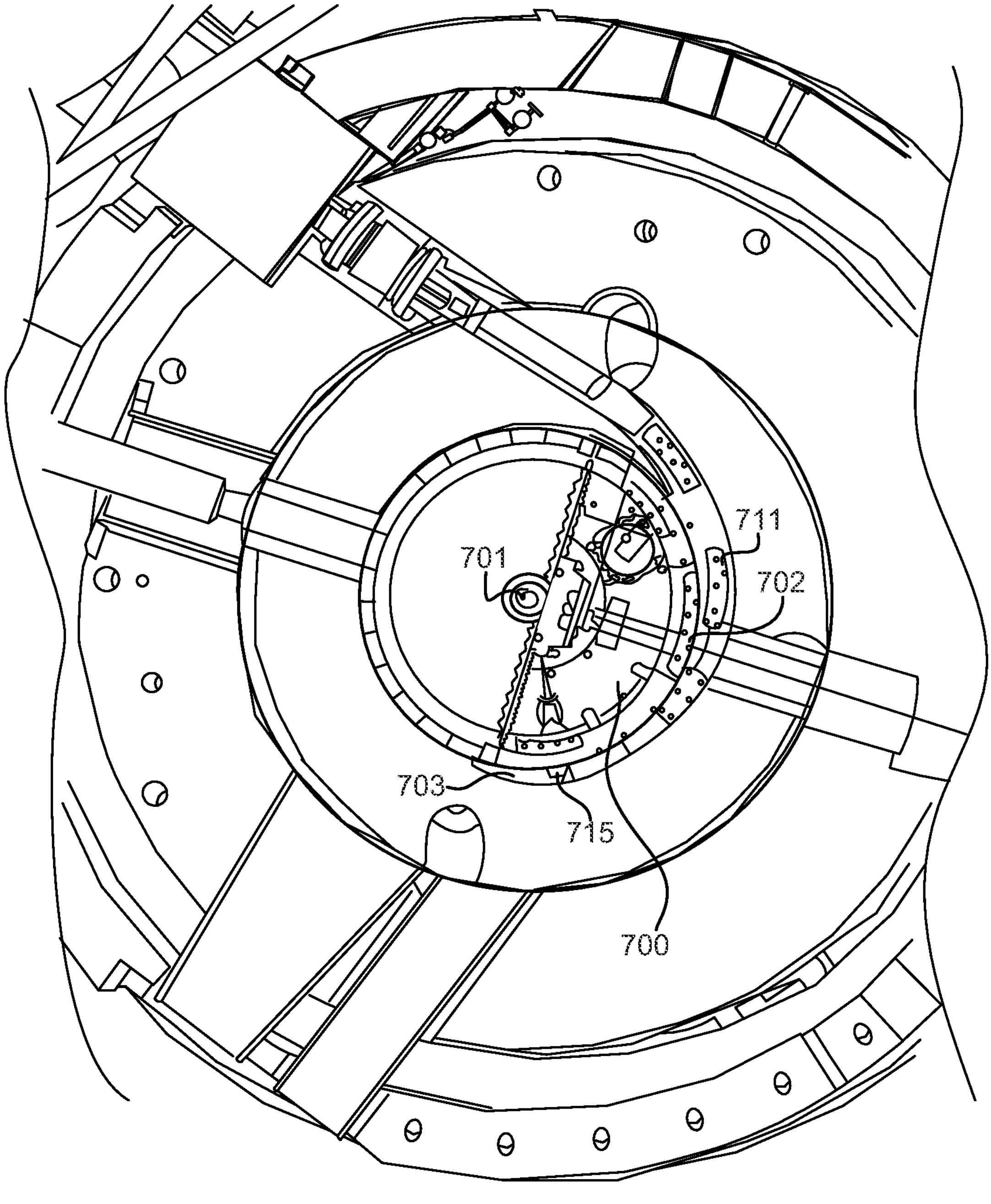

FIG. 15

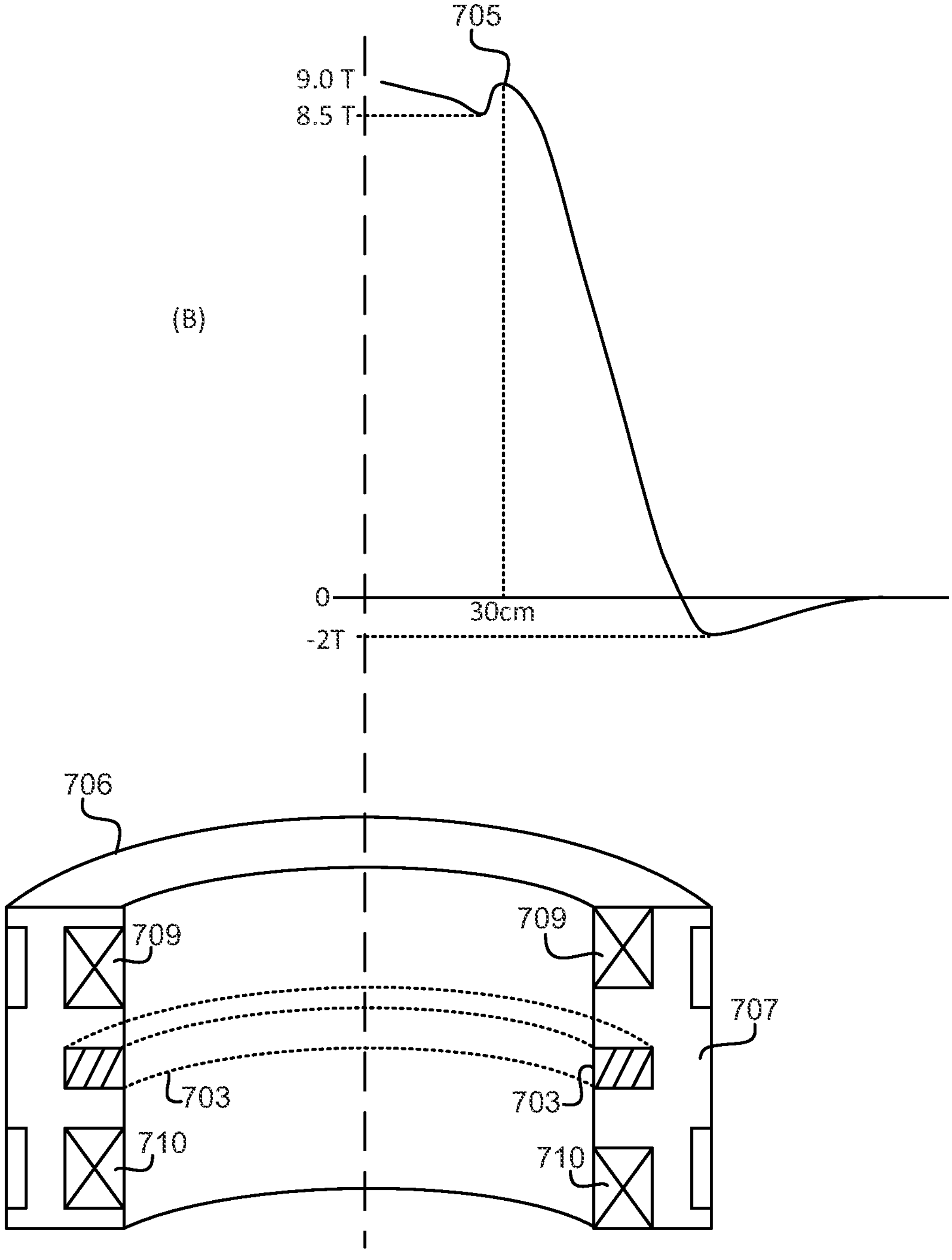

FIG. 16

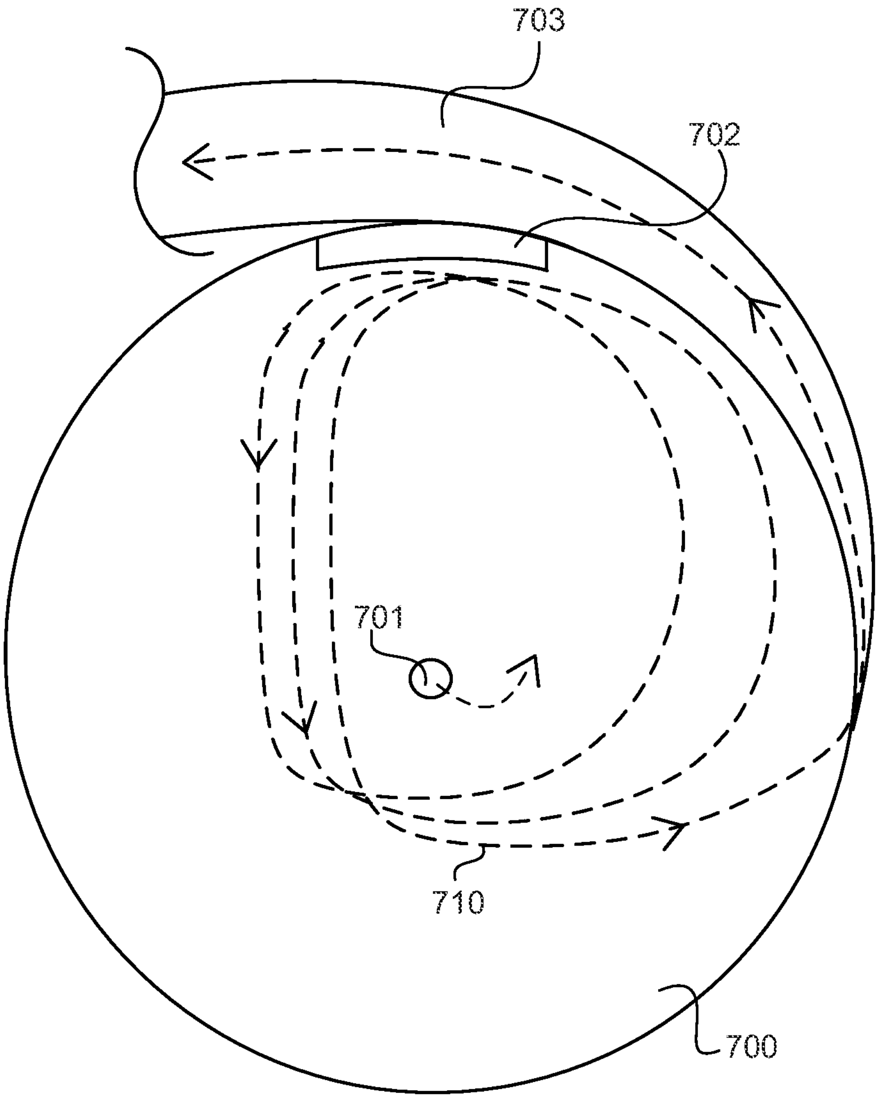

FIG. 17

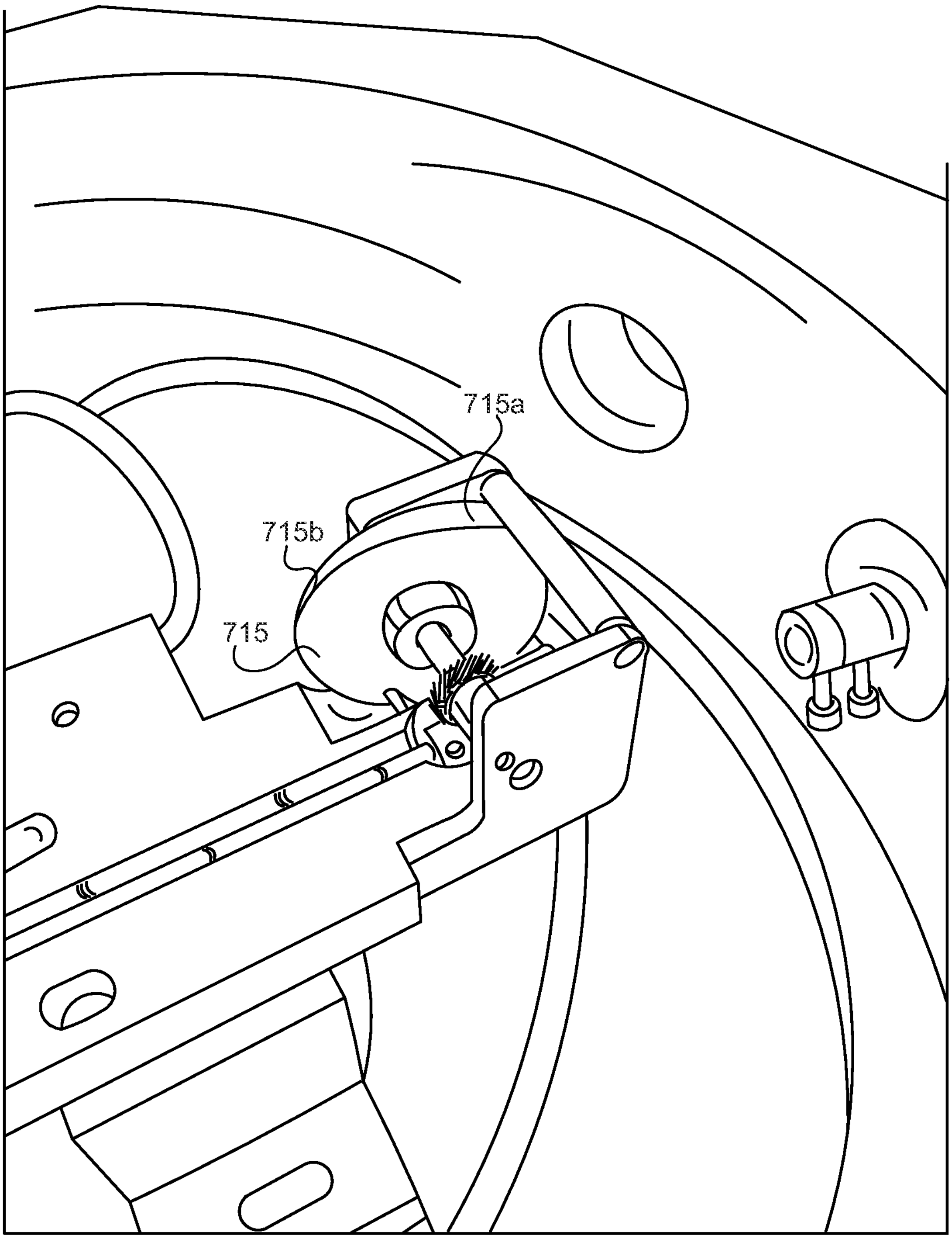

FIG. 18

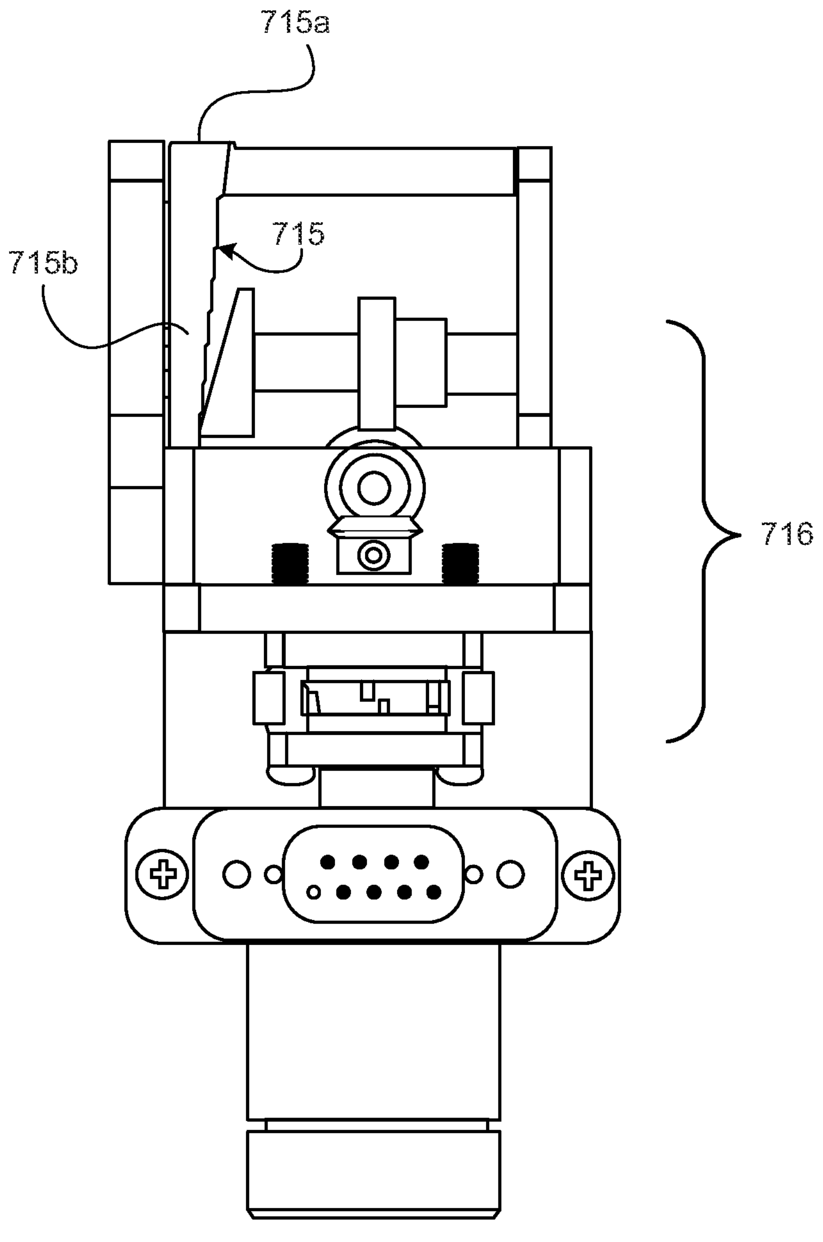

FIG. 18A

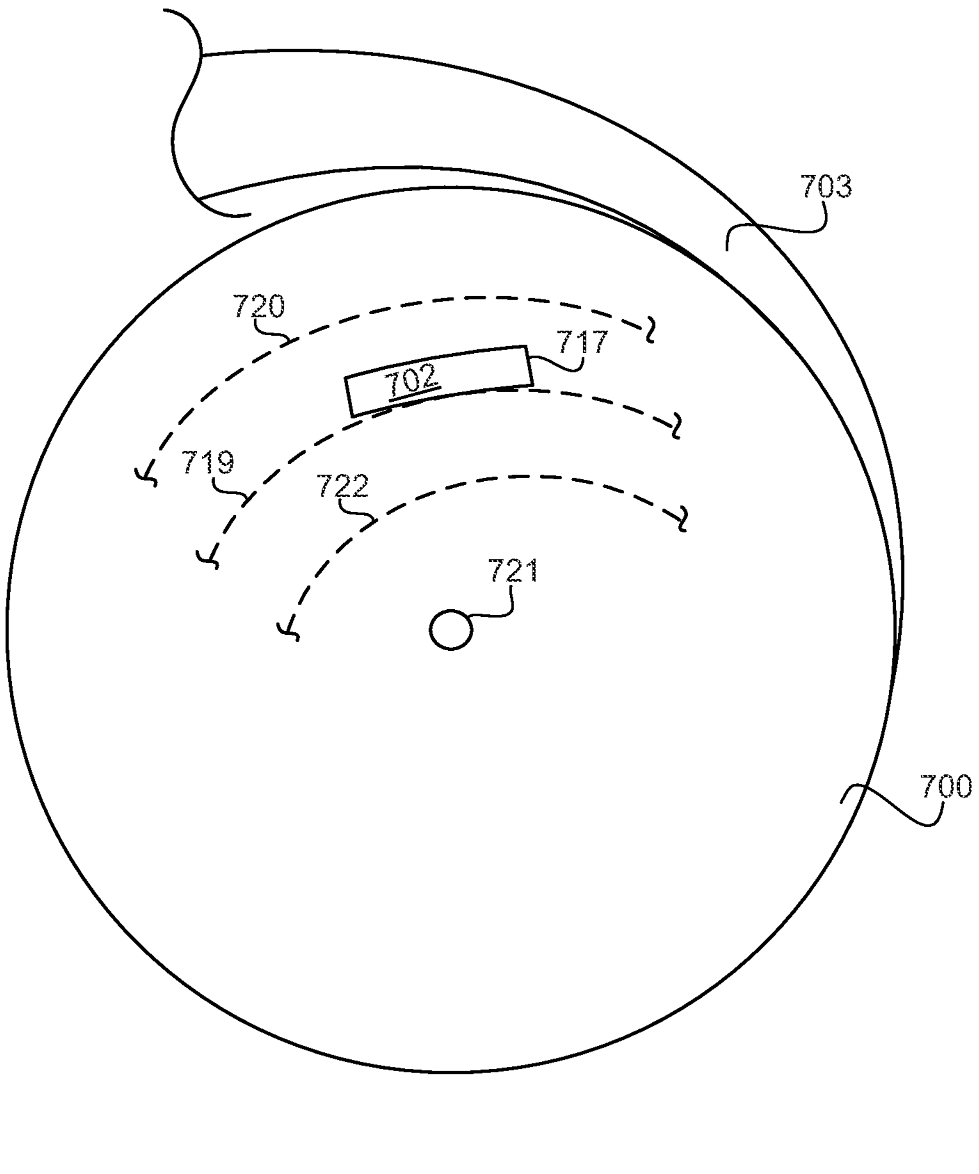

FIG. 19

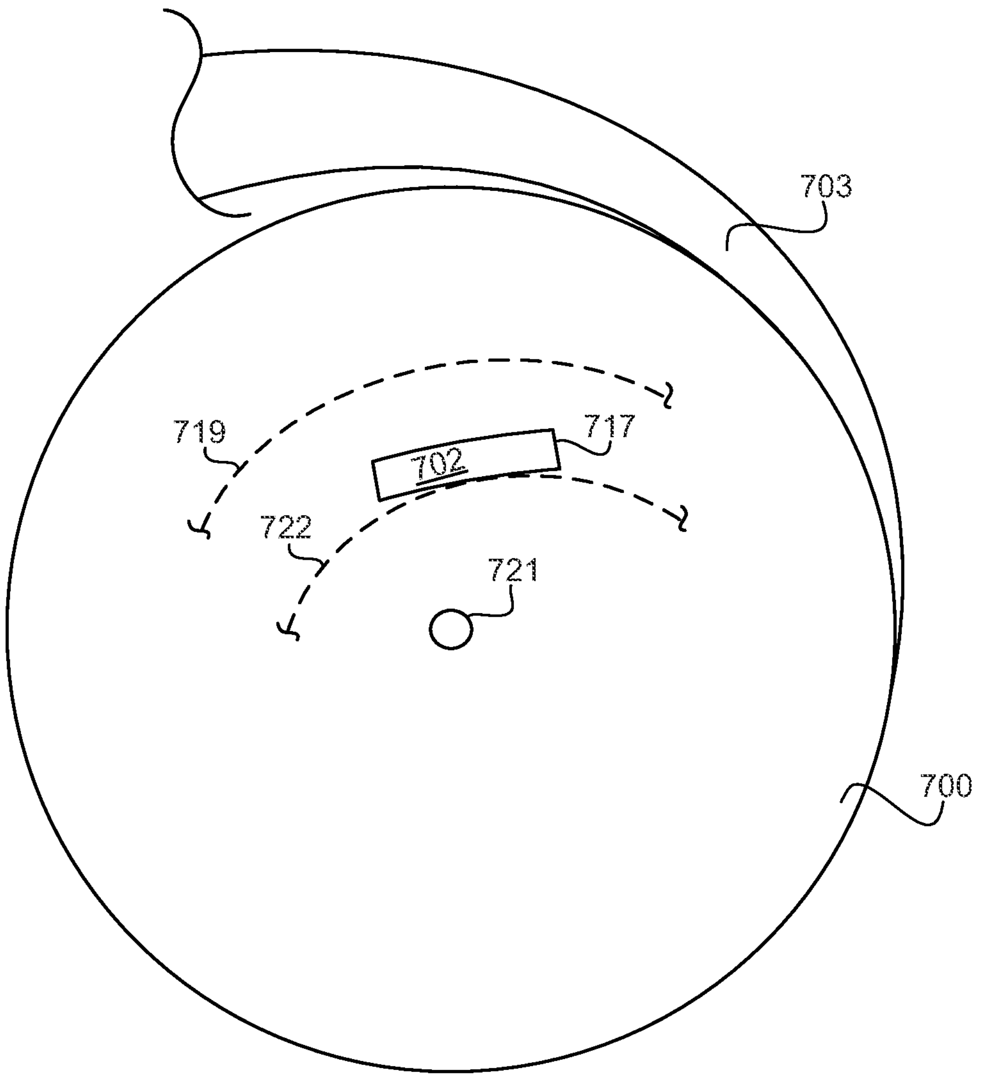

FIG. 20

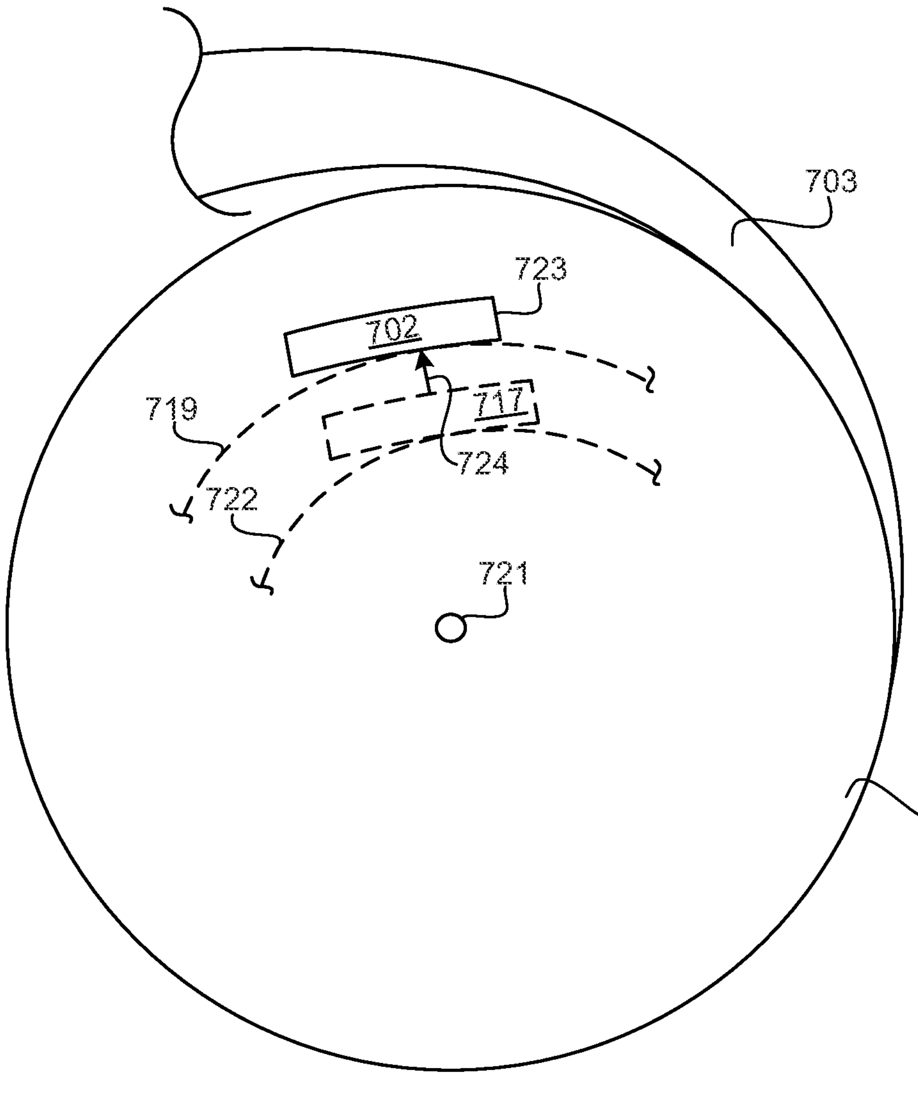

FIG. 21

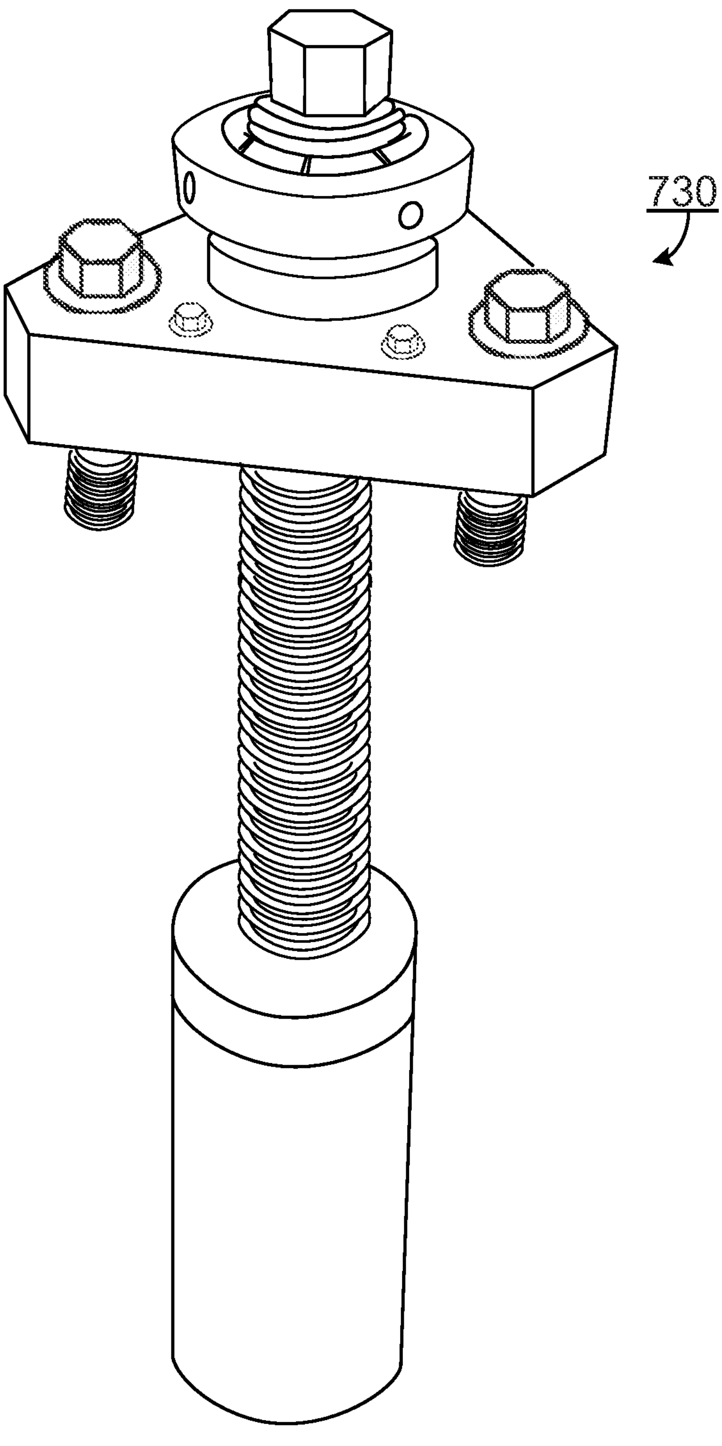

FIG. 22

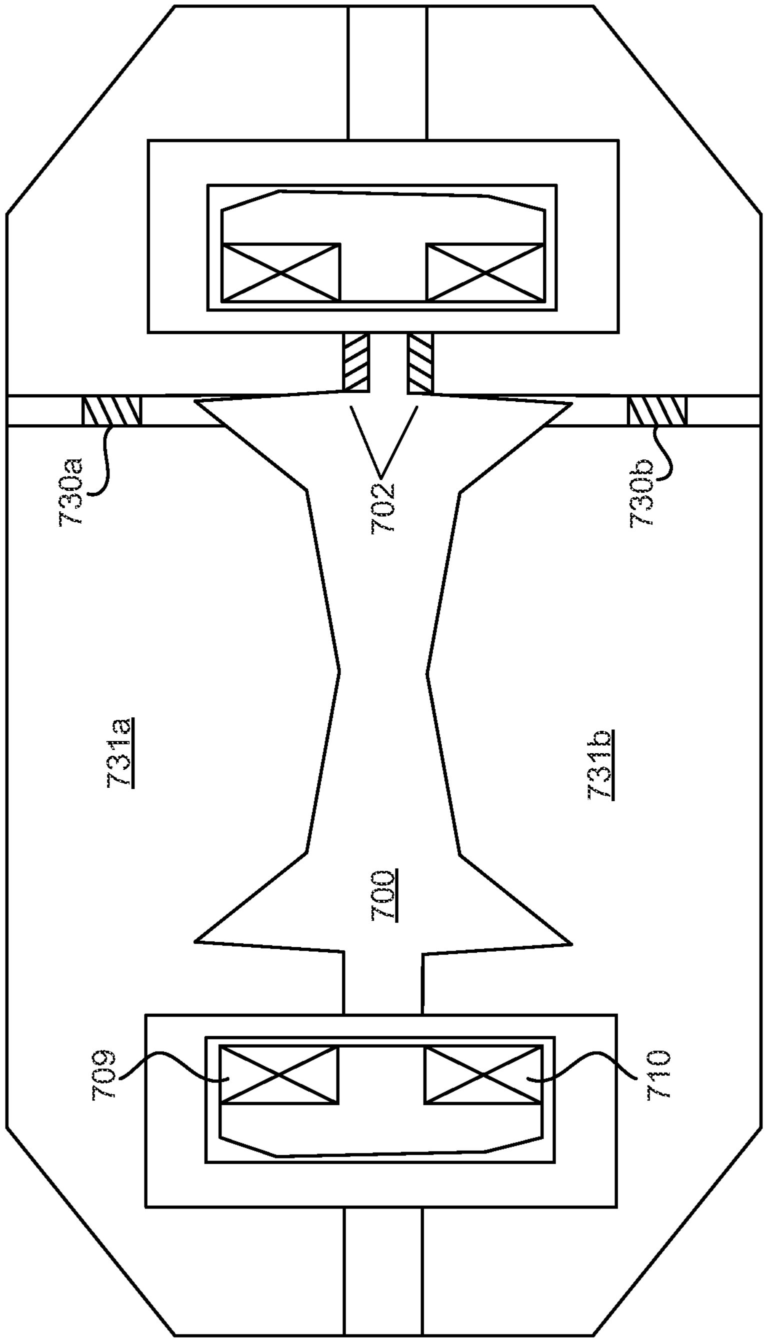

FIG. 23

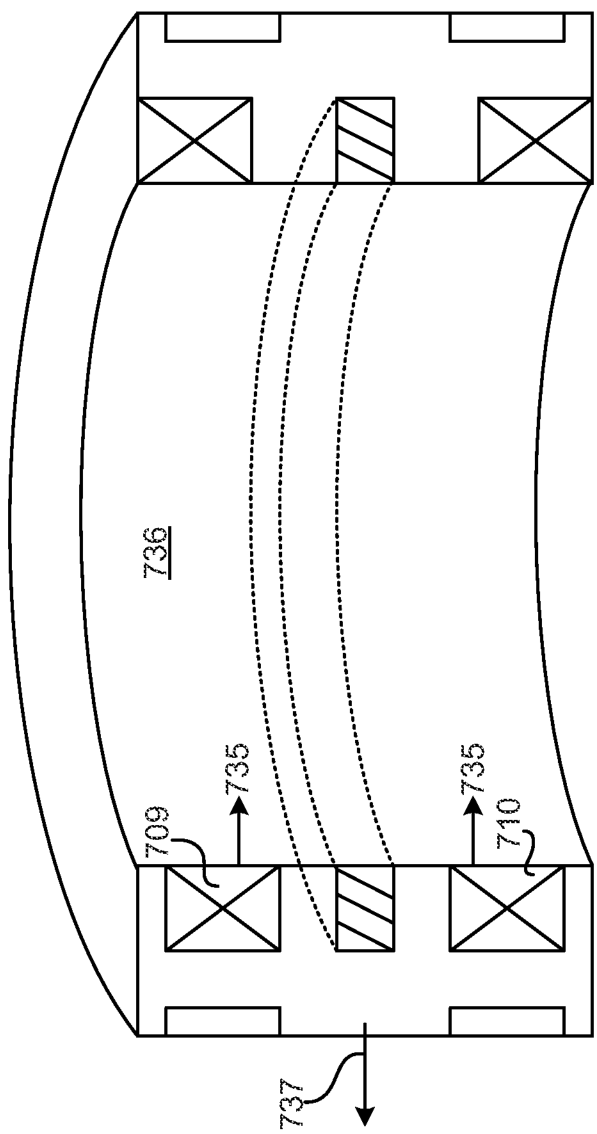

FIG. 24

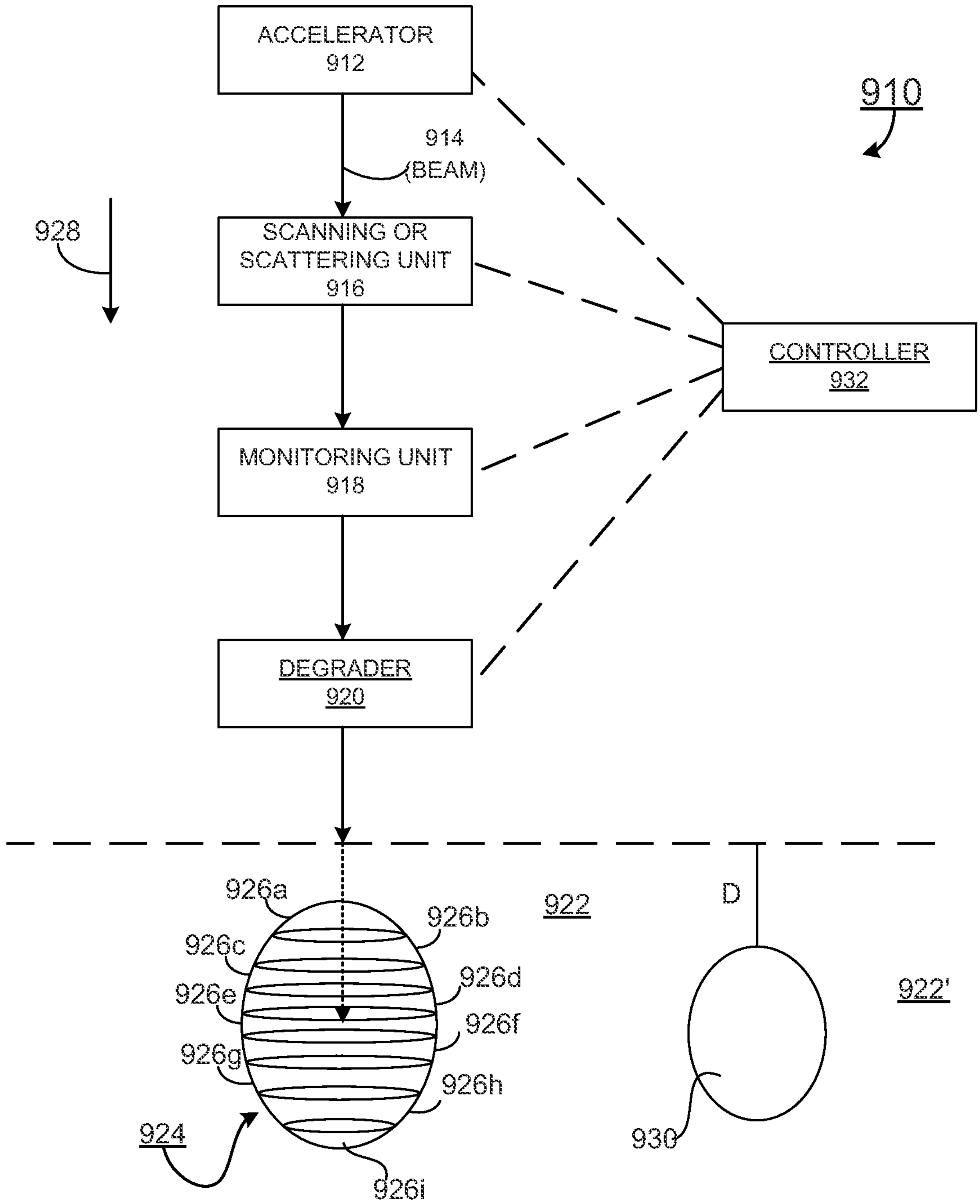
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FIG. 25

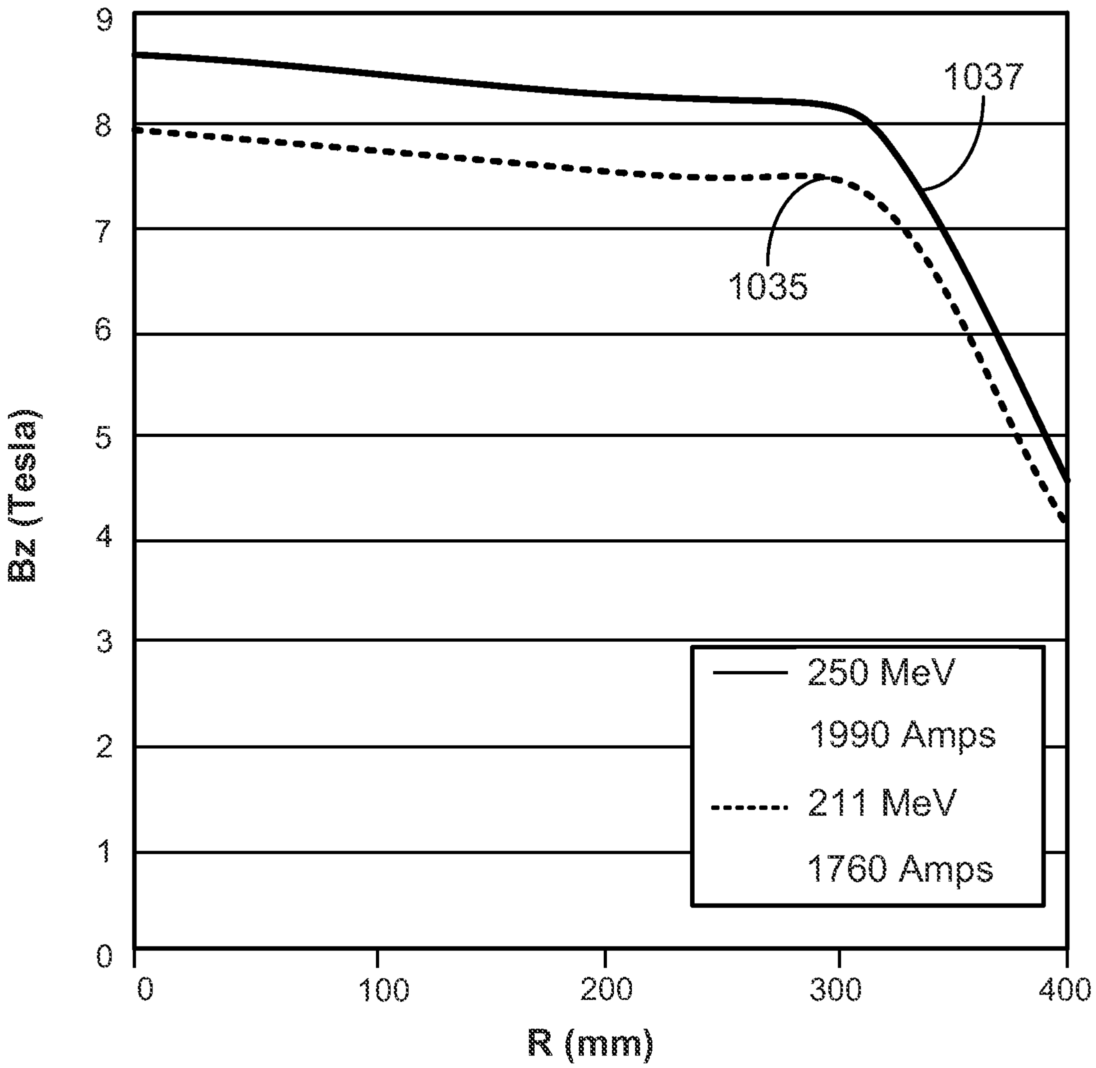
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FIG. 26

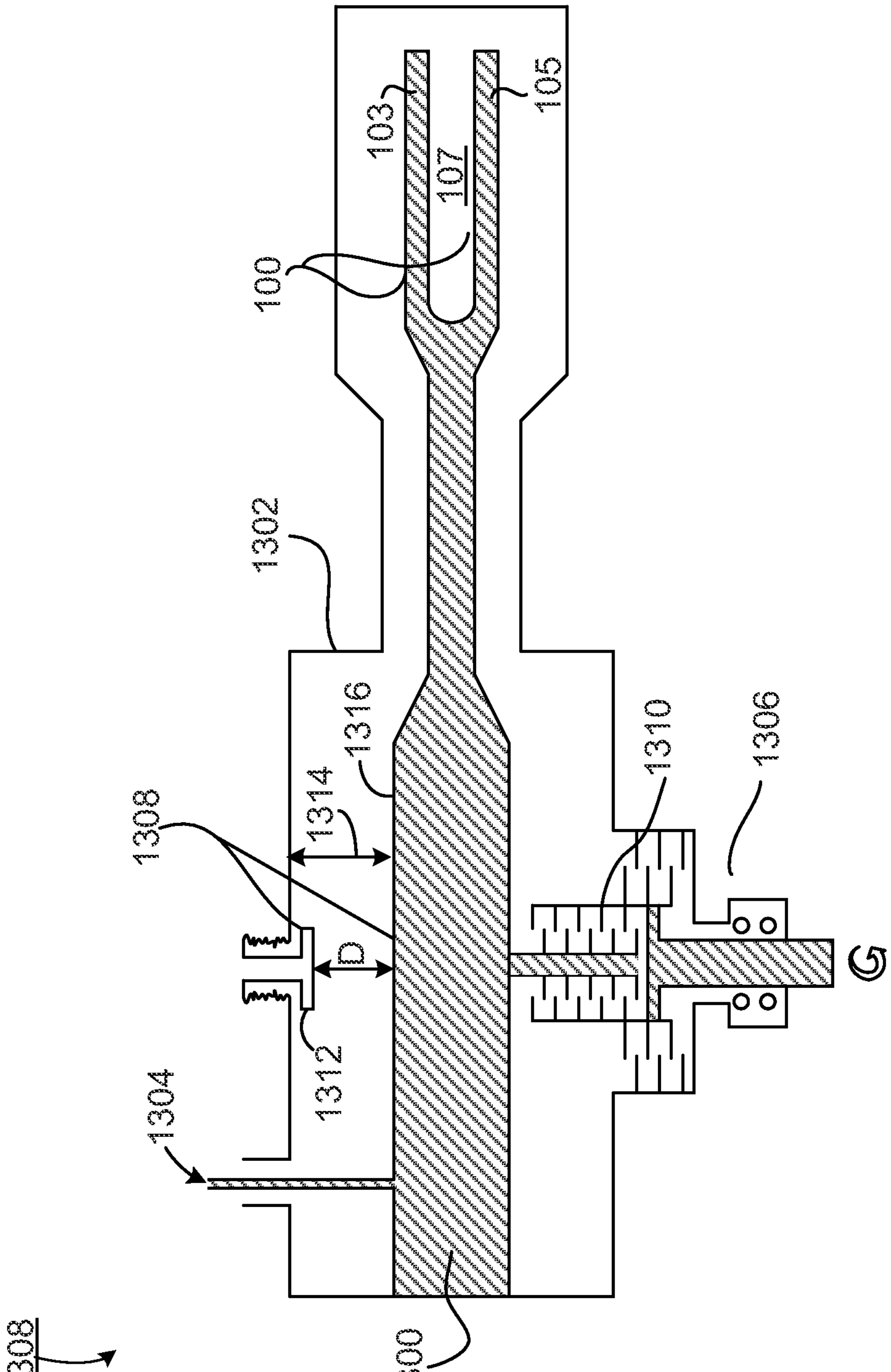
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FIG. 27

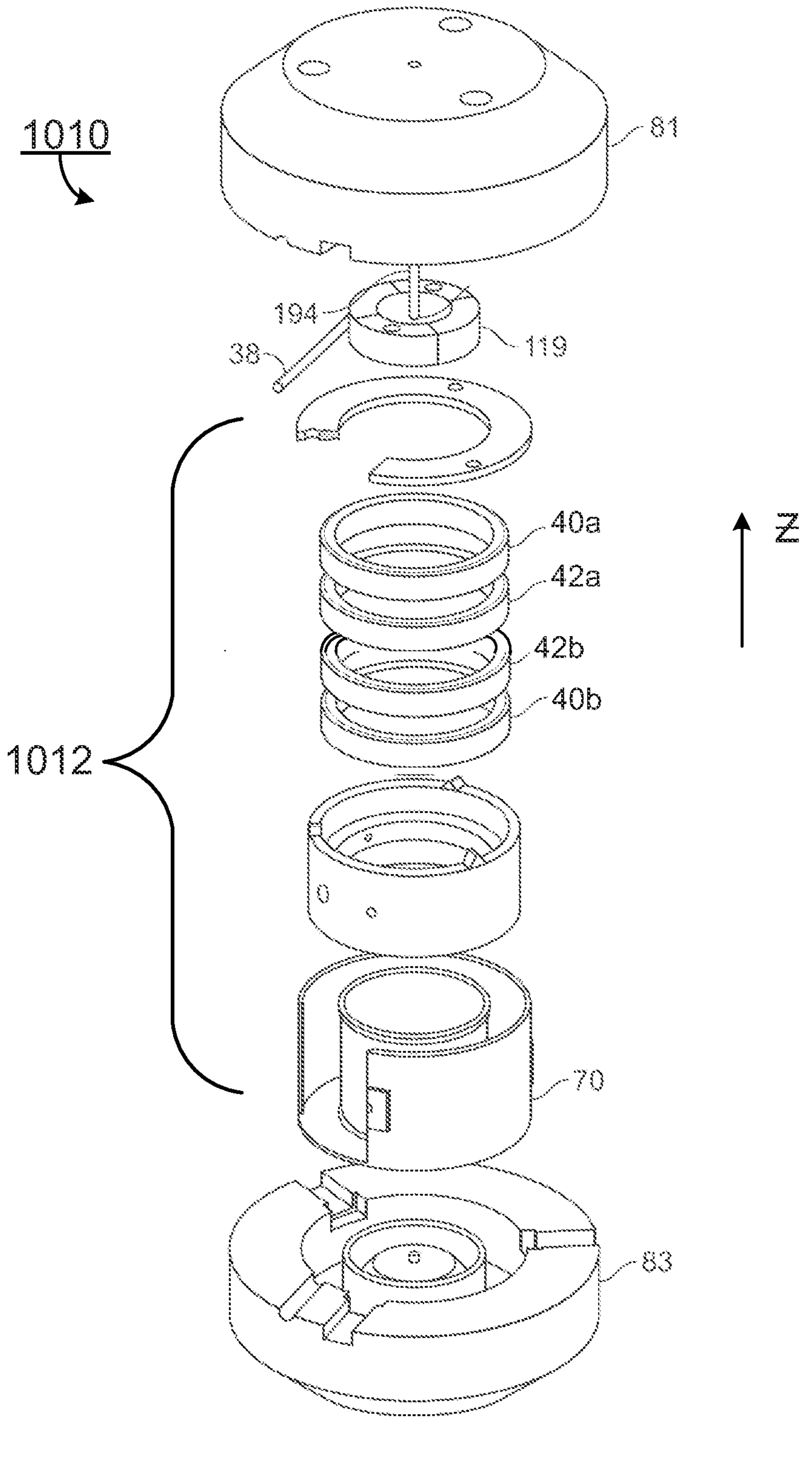

FIG. 28