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

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
23 August 2001 (23.08.2001)

PCT

(10) International Publication Number
WO 01/61724 A1

(51) International Patent Classification[7]: H01J 35/00

(21) International Application Number: PCT/US01/05359

(22) International Filing Date: 16 February 2001 (16.02.2001)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
09/504,709 16 February 2000 (16.02.2000) US
09/566,662 8 May 2000 (08.05.2000) US

(71) Applicant: X-TECHNOLOGIES, LTD. [US/US]; 40 Ramland Road, South, Suite 10, Orangeburg, NY, 10962-2617 (US).

(74) Agent: ROSSI, Marc, A.; Rossi & Associates, P.O. Box 826, Ashburn, VA 20146-0826 (US).

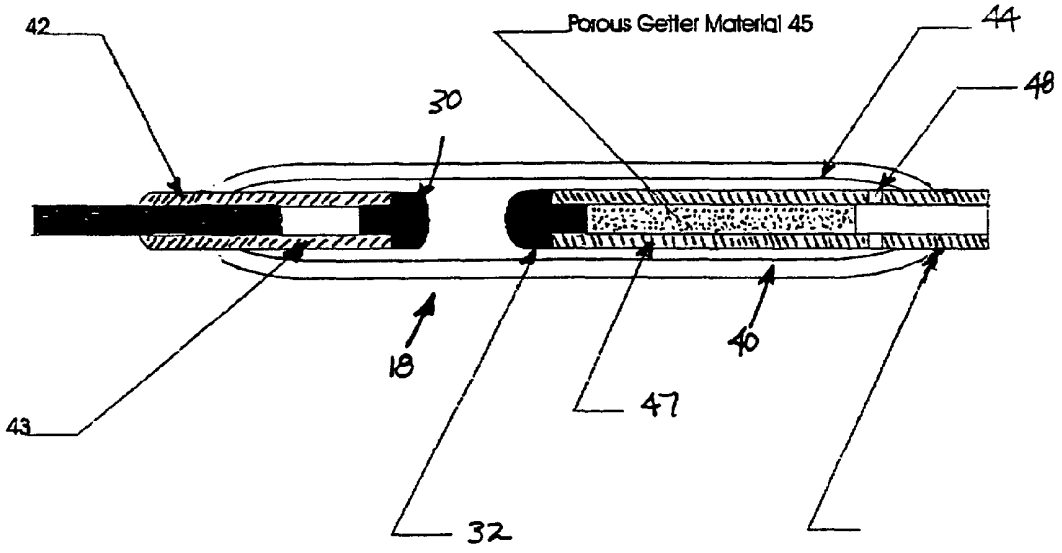
(81) Designated States *(national)*: AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BY, BZ, CA, CH, CN, CR, CU, CZ, DE, DK, DM, DZ, EE, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KP, KR, KZ, LC, LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW, MX, MZ, NO, NZ, PL, PT, RO, RU, SD, SE, SG, SI, SK, SL, TJ, TM, TR, TT, TZ, UA, UG, UZ, VN, YU, ZA, ZW.

(84) Designated States *(regional)*: ARIPO patent (GH, GM, KE, LS, MW, MZ, SD, SL, SZ, TZ, UG, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, CY, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE, TR), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, GW, ML, MR, NE, SN, TD, TG).

Published:
— *with international search report*
— *before the expiration of the time limit for amending the claims and to be republished in the event of receipt of amendments*

For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.

(54) Title: MINIATURE ENERGY TRANSDUCER FOR EMITTING X-RAY RADIATION



(57) Abstract: An apparatus and method for providing in-situ radiation treatment in humans includes a miniature energy transducer to emit x-rays. The energy transducer includes a transducer body (40), a cathode (30) provided at one end of the transducer body, and an anode (32) provided at another end of the transducer body. The transducer body, cathode and anode define a cavity in which a desired vacuum is maintained by getter material (45). In a preferred embodiment an explosive emission cathode is utilized. The energy transducer has an outer diameter and length sufficient to enable use in human arterial systems for the treatment of restenosis and other conditions.

WO 01/61724 A1

MINIATURE ENERGY TRANSDUCER FOR EMITTING X-RAY RADIATION

Cross-Reference to Related Applications

This application is a continuation-in-part application of, and claims priority from, U.S. Patent Application Serial Number 09/434,958, the contents of which are incorporated herein by reference, which in turn is a continuation-in-part application of U.S. Patent Application Serial Number 09/325,703 filed June 3, 1999, the contents of which are incorporated herein by reference, which in turn claims priority from Provisional Patent Application Serial Number 60/087,970 filed June 4, 1998. This application is also a continuation-in-part application of, and claims priority from, U.S. Patent Application Serial Number 09/504,709, the contents of which are incorporated herein by reference, filed February 16, 2000.

Field of the Invention

The present invention relates to an apparatus and method for providing x-ray brachytherapy treatment in humans. More specifically, the present invention relates to an apparatus and method for providing in-situ radiation treating that utilizes a miniature energy transducer to produce x-rays, wherein the energy transducer includes a getter material, applied in various forms, to maintain a desired vacuum level.

Background of the Invention

Restenosis is a heart condition that afflicts 35%-50% of all people who undergo balloon angioplasty to improve blood flow in narrowed sclerotic arteries. The condition consists of a significant re-closing of the treated artery segment hours to several months after the procedure. As a result, the arterial lumen size is decreased and the blood flow downstream from the lesion site is impaired. Consequently, patients afflicted with restenosis must undergo an additional balloon angioplasty, and in some cases a coronary bypass surgery must be performed. Aside from the pain and suffering of these patients, recurrent stenosis is also a serious economic burden on society, with estimated expenses as high as 3.0 billion dollars per year in the United States economy alone.

Attempts to treat restenosis have been concentrated in both the pharmacological and medical device areas. While pharmacological solutions have been proven effective in treating only acute restenosis, a condition developing immediately after balloon angioplasty, some progress has been made with medical devices in the treatment of long term restenosis, a condition that develops up to a few months following balloon angioplasty. An example for such medical device is the stent. Stents can be inserted into an occluded artery to hold it open. Stents have been shown to prevent two of the three mechanisms that cause recurrent stenosis, namely, elastic recoil of the artery and negative remodeling of the arterial structure. The third mechanism, neointimal growth, consists of hyper-proliferation of smooth muscle cells from the lesion into the lumen and is not prevented by stents.

Ionizing radiation holds great promise for treating restenosis. Ionizing radiation serves to damage undesirable hyper-proliferating tissue and ultimately to prevent the hyper-proliferation of smooth muscle cells in the irradiated region. Research has shown that gamma and beta radiation delivered at the location of stenotic lesions effectively stop both animal and human intimal proliferation. The effective, yet non-hazardous, required dose to treat human restenosis is between seven and forty Gray (mjoule/gram), preferably a dosage greater than fifteen Gray, that penetrates the artery wall at a two mm depth over the lesion length.

In view of the above, various methods have been proposed to provide ionizing radiation treatment. For example, radiation catheters, based on the use of radioactive sources such as beta - emitting ^{32}P, $^{90}Sr/^{90}Y$, $^{188}W/^{188}Re$, beta+ emitting ^{48}V or gamma emitting ^{192}Ir, are at various stages of development and clinical evaluation. The radioactive sources, in a variety of configurations, are introduced to the treatment sites using special radiation catheters and the radioactive source is placed at the treatment site for a predetermined time period as to deliver the proper radiation dose. Radioactive stents are also used as alternative delivery means, composed of the above radioactive isotopes.

The gamma and beta radioactive sources used by the present radiation catheters and radioactive stents, however, have several drawbacks including a limited ability to provide selective control over the dose distribution or overall radiation intensity, and the logistical, regulatory, and procedural difficulties involved in dealing with radioactive materials. In addition, gamma-emitting devices jeopardize patients by exposing healthy organs to dangerous radiation during the introduction of the radiation source. Hospital personnel that

handle radioactive materials are also at risk due to exposure. In addition to the risks these devices impose on patients, hospital staff, and the environment, use of these devices invokes a regulatory burden due to the need to comply with nuclear regulatory requirements.

An additional approach to providing ionizing radiation treatment is through the use of an x-ray emitting energy transducer, which is not radioactive. Conventional x-ray radiation for radiotherapy is produced by high-energy electrons generated and accelerated in a vacuum to impact a metal target. The x-ray emission is directly proportional to the electron beam current. However, the efficiency of x-ray generation is independent of electron current, but rather depends on the atomic number of the target material and on the acceleration voltage. Yet, another method for the production of x-rays is by direct conversion of light into x-ray radiation. It is known that the interaction of light with a target can produce highly energetic x-rays when the power densities achieved are in the range of 10^{16}-10^{17} watt/cm^2. With the development of femtosecond laser, such power densities are achievable with moderate size lasers (See C. Tillman et al, NIMS in Phys. Res. A394 (1997), 387-396 and US Patent No. 5,606,588 issued to Umstadter et al., the contents of each of which are incorporated herein by reference). A 100 femtosecond , one mJ laser pulse focused down to a 3 micron spot, for example, will reach this power density level.

A variety of medical applications of the direct laser light conversion method of x-ray generation are currently in the development stage. The direct laser light conversion method, for example, has been considered for medical imaging (See, Herrlin K et al.

Radiology (USA), vol. 189, no. 1, pp. 65-8, Oct. 1993). Another medical application of femtosecond lasers is in improved non-thermal ablation of neural or eye tissue for surgical purposes (See, F.H. Loesel et al. Appl.Phys.B 66,121-128 (1998)). The development of compact table top models of femtosecond lasers makes laser generated x-rays an attractive alternative for radioactive material based radiotherapy.

Based on the above, an x-ray brachytherapy treatment apparatus and method has been developed. In x-ray brachytherapy an internal x-ray emitting miniature energy transducer generates x-rays in-situ. Co-pending and commonly assigned U.S. Patent Application Serial Number 09/325,703 filed June 3, 1999, and U.S. Patent Application Serial Number 09/434,958 filed November 5, 1999, describe miniaturized energy transducers that are coupled to flexible insertion devices to permit x-ray radiation treatment within the human body. Use of the miniaturized x-ray emitting energy transducer offers certain advantages with respect to intra vascular gamma and beta sources. These advantages are, but are not limited to, localization of radiation to the treatment site so that the treatment site may be irradiated with minimal damage to surrounding healthy tissue; reduction of hospital personnel risk due to exposure to radioactive materials; and minimization of the regulatory burden and additional costs that arise from the need to comply with nuclear regulatory requirements.

A variety of different types of cathode and anode structures have been proposed for the energy transducer. One proposal is to utilize a hollow cathode, which includes a cathode shell that defines a cavity. A light pulse is introduced into the cavity in order to heat an outer surface of the cathode shell, thereby causing thermionic emission of electrons

from the outer surface. Another proposal for a hollow cathode incorporates the use of an electron escape nozzle, wherein an ion and electron plasma is generated in the cavity either by applying a light signal to an inner surface of the cathode shell or by providing a spark gap in the cavity of the conducting cathode shell. The electrons exit the cathode shell via the escape nozzle and are accelerated to the anode upon the application of a voltage pulse to the cathode. Still further, in a linear reverse cathode emission type of transducer, an anode is located at a first end of a transducer body and an emission element is located at a second end of the transducer body opposite the anode. The emission element is either a photo-emission electron source or a thermionic emission surface, and generates electrons when activated by a light source. Furthermore, in a distally illuminated plasma cathode an anode is located at the first end of a transducer body and a cathode is located at a second end of a transducer body opposite to the anode. An ion and electron plasma is generated by applying a light pulse to the surface of the cathode. Electrons are extracted from the plasma by the application of a high voltage pulse to the anode. They are accelerated by the said high voltage pulse and strike the anode, where x-rays are produced.

Co-pending and commonly assigned U.S. Patent Application Serial Number 09/504,709, filed February 16, 2000, the contents of which are incorporated herein by reference, describes an explosive electron emission plasma cathode structure wherein one of the preferred embodiments utilizes liquid metal such as gallium-indium. The liquid metal is deposited inside a hollow, needle shaped, conducting cathode, which is made, for example, out of tungsten. A pre-pulse of voltage, typically 1-10 kV in magnitude, is applied to the cathode for duration of less than 100 microseconds. As a result, the surface

of the liquid metal obtains the shape of a sharp tip having a typical diameter of 1-20 microns. A second pulse is then applied, typically of 10-100 kV, for duration of 1-100 nanoseconds. As a result, plasma is formed at the liquid metal tip. The high voltage pulse extracts electrons out of the plasma and accelerates them towards the anode. At the anode, the electrons' kinetic energy is converted to x-ray photons and heat. Further still, in another preferred embodiment of said plasma cathode, the cathode may be an explosive electron emission cathode that contains a single carbon fiber or a bundle of carbon fibers. A high voltage pulse, of typical magnitude of 10-100 kV and typical duration of 1-100 nanoseconds, is applied to the cathode. As a result, the carbon fiber or fibers' tips are sublimated and plasma is formed. The high voltage pulse extracts electrons out of the plasma. These electrons are accelerated by the high voltage pulse towards the anode. At the anode, the electrons' kinetic energy is converted to x-ray photons and heat.

Regardless of the type of anode and cathode structure utilized, an appropriate level of vacuum must be maintained within the transducer body. A vacuum is required to permit the production of electrons at the cathode and the emission of electrons from the transducer body, to enable uninterrupted acceleration of electrons from the cathode to the anode over a voltage difference that can typically vary from 10 kV to 100 kV, and to prevent high voltage breakdown within the transducer body, as breakdown probability is enhanced by ionization of the molecules of any residual gases present within the transducer body. The required vacuum level depends mainly on the electron production mechanism, which is employed in the device. This level can vary from 10^{-10} Torr for x-ray emitters utilizing surface based electron emission mechanisms such as photo-emission or field emission,

down to 10^{-2} Torr or less for x-ray emitters employing pulsed plasma based mechanisms. The required evacuated volume is about 1 mm^3 as the transducer body is preferably several millimeters in length and about 1 mm in diameter. If a typical leak rate of gas into the cavity is 10^{-11} Torr-liter/sec, the pressure inside the cavity will rise at a rate of approximately 1 Torr/day. Unless a mechanism exists to absorb the entrant gases, the vacuum level will be quickly degraded to a point where x-ray emission is not possible.

The above-mentioned analysis also applies to disorption of gases during the operation of the energy transducer. The surface area defining a cavity within the transducer body is about 10 mm^2. Thus, the ratio of surface area to volume is approximately 10 mm^{-1}. The length and diameter of conventional x-ray tubes sized for most medical applications are about 100 times larger than the miniaturized energy transducer being considered, and conversely, the surface to volume ratio is 100 times smaller than that of the miniaturized energy transducer. Out-gassing is a major potential source of vacuum degradation. For any given surface specific out-gassing rate, the above mentioned comparison implies that the vacuum will be degraded 100 times faster in a miniaturized energy transducer than a conventional x-ray tube. Thus, enhancing vacuum preservation before and throughout the operation of an x-ray emitting miniaturized energy transducer is particularly crucial.

U.S. Patent 5,854,822 and PCT Patent Application WO97/077040 to Chornenky, et al describe a field emission cathode wherein the cathode is made from getter material. Although using getter as a cathode enables the design of a relatively small x-ray emitter, this combination has certain disadvantages. Reproducibility and stability of the emitted

current, which is especially crucial to the production of the precise required x-ray dose needed for the treatment of restenosis, cannot be achieved due to surface irregularities, unless conditioning of the cathode takes place. The electric fields at the surface of such cathodes have to be kept as weak as possible to avoid the risk of electrical flashover enhanced by cathode surface imperfections. Thus it is an object of the present invention to provide a mechanism for preserving the vacuum level in miniaturized energy transducers that emit x-rays without interfering with electron emission at the cathode or x-ray emission at the anode and without effecting the properties of the emitted current and the parameteres of the x-ray emitter.

In view of the above, it is another object of the present invention to provide a simple, inexpensive method to maintain vacuum level within a miniature x-ray emitter that does not require a conditioning process prior to the assembly of the x-ray emitter.

Another object of the present invention is to utilize the heat generated at the anode during x-ray production, conducted through the getter material, to constantly activate the getter material. During this heat activation process, the getter surface layer that absorbs stray gas molecules is decomposed and a fresh, ultra-clean surface of getter layer is exposed, enabling enhanced vacuum preservation.

Summary of the Invention

The invention provides an apparatus and method for x-ray brachytherapy treatment in humans, wherein different types of miniature energy transducers utilized to emit x-rays.

More specifically, in a preferred embodiment, the energy transducer includes a transducer body, a cathode provided at one end of the transducer body, and an anode provided at another end of the transducer body. The transducer body, cathode and anode define a cavity in which a desired vacuum is maintained by a getter material provided in various configurations, for example, utilizing the structural elements of a generic structure made from glass and a low expansion glass binding metal or metal alloy. The getter material is preferably titanium, zirconium, aluminum, iron, vanadium, tantalum or an alloy of at least two of these materials. In some preferred embodiments the getter material absorbs the thermal energy, which is created during x-ray production at the anode. Such thermal energy increases the temperature of the getter, thus causing its reactivation during operation by a mechanism involving the diffusion of sorbed gases into the bulk of the getter material and the exposure of a fresh, ultra-clean surface of getter. The energy transducer has an outer diameter of less than 2 mm and a length of less than 10 mm, thereby enabling the device to be utilized in human arterial systems for the treatment of restenosis and other conditions.

Other advantages and features of the invention will become apparent from the following detailed description of the preferred embodiments and the accompanying drawings.

Brief Description of the Drawings

The invention will now be described with reference to certain preferred embodiments thereof and the accompanying drawings, wherein:

Fig. 1 is a functional block diagram of an x-ray brachytherapy system in accordance

with the present invention;

Fig. 2 is a cross-sectional view of an energy transducer in accordance with the invention in a linear configuration;

Fig. 3 is a cross-sectional view of an energy transducer in accordance with the invention in a linear reverse configuration;

Fig. 4 is a cross-sectional view of an energy transducer that includes getter material confined within an anode trunk;

Fig. 5 is a cross-sectional view of an energy transducer that includes getter material surrounding the neck of a conducting anode;

Fig. 6 is a cross-sectional view of an energy transducer that includes getter material in the form of a getter trunk;

Fig. 7 is a cross-sectional view of an energy transducer that includes a getter coating on the surface of an anode trunk;

Fig. 8 is a cross-sectional view of an energy transducer that includes a laser induced plasma cathode;

Fig. 9 is a cross-sectional view of an energy transducer that includes a laser induced thermionic cathode;

Fig. 10 is a cross-sectional view of an energy transducer that includes a distal illumination laser cathode;

Fig. 11 is a cross-sectional view of another embodiment of a laser induced thermionic/photo-emissive cathode;

Fig. 12 is a cross-sectional view of an energy transducer that includes a spark induced

plasma cathode;

Fig. 13 is a cross-sectional view of a tip of an explosive electron emission plasma cathode that includes a liquid metal cathode; and

Fig. 14 is a cross-sectional view of a tip of an explosive electron emission plasma cathode that includes a carbon fiber cathode.

Detailed Description of the Preferred Embodiments

The present invention is particularly applicable to a system for the delivery of x-ray radiation to localized targets inside and outside the human body. Some therapeutic uses for the invention include the irradiation of coronary lesions to prevent restenosis, the irradiation of tumors, and arterio-venous malformations. It will be understood, however, that the invention is not limited to these particular applications.

Fig. 1 illustrates an x-ray brachytherapy system 10 in accordance with the present invention. The system 10 includes a miniature energy transducer 18 that is coupled to a distal end of a flexible insertion device 12. The flexible insertion device 12 delivers energy from an energy source 14 to the miniature energy transducer 18, which preferably converts electrical and/or optical signals received from the energy source 14 into x-ray radiation and distributes the x-ray photons (illustrated by arrows 20) in a predetermined distribution pattern. The energy source 14 is preferably located external to the patient, while the flexible insertion device 12 is manipulated to place the miniature energy transducer 18 in an area to be treated within the patient's body.

The miniature energy transducer 18 is preferably surrounded by x-ray transmissive insulation (not shown) that can be presented in direct contact with the human body. The transmissive insulation may be a material coated on an outer surface of the miniature energy transducer 18. Alternatively, the transmissive insulation may take the form of a capsule that encapsulates the miniature energy transducer 18. In any case, the miniature energy transducer 18 is preferably a relatively low-cost, replaceable and disposable unit, which avoids the necessity of complex sterilization processes required for instruments that are intended for multiple uses.

The system 10 may further include a dosimetery system 22 incorporating a dosimeter measurement unit 24 connected to a scintillating optical fiber 26. Preferably, the scintillating optical fiber 26 is a standard plastic scintillating optical fiber, containing embedded dopant atoms, which produce light photons upon being irradiated with x-ray photons. The distal end of the scintillating optical fiber 26 is located in the immediate vicinity of the miniaturized energy transducer 18. Preferably, the bulk of the length of the scintillating optical fiber 26 is housed within the flexible insertion device 12. The optional dosimeter measurement unit 24 is preferably housed within the energy source 14, and it is connected to a control unit 28, which is also housed within the energy source 14.

In general, the energy source 14 is adapted to provide electrical and/or optical signals through the flexible insertion device 12 that is correspondingly configured to deliver the energy to the miniature energy transducer 18. Accordingly, the energy source 14 is provided with a power supply 11, such as voltage pulse generator, and/or a light source 15, for example a laser, respectively connected through an electrical conductor 17,

preferably a coaxial cable, and an optical conductor 19 to the flexible insertion device 12. The control unit 28 directs the energy source 14 to deliver electrical and/or optical signals through the flexible insertion device 12 to the miniature energy transducer 18 as required by the operator. During x-ray treatment, the control unit 28 may also receive information from the optional dosimeter measurement unit 24 and may use this information to achieve the required dosage amount of radiation, providing the system with a feedback mechanism. Thus, the duration and amplitude of the energy supplied by the power supply 11 and optionally by the light source 15, as well as the total treatment time, may be varied to control the distribution of the x-ray radiation produced by the miniature energy transducer 18.

The flexible insertion device 12 preferably combines the electrical and the optional optical transmission lines into one structure, thus economizing the catheter diameter that must be preferably 1.7 mm or less, for coronary applications, and thereby allowing the flexible insertion device 12 to follow the contours of a blood vessel or any other body cavity.

Referring now to Fig. 2, a preferred embodiment of the energy transducer 18 in a linear configuration is disclosed. The energy transducer 18 includes a conducting cathode 30 and a conducting anode 32 respectively located at a proximal end and a distal end of a transducer body, which in the illustrated embodiment is an electrically insulating tube 34. The conducting cathode 30 optionally includes a cathode tip 37, depending on the type of energy transducer that is being used. The sealed tube 34 is preferably 3-10 mm in length and 0.5-1.7 mm in diameter, and holds a vacuum within a cavity 38 that typically varies from 10^{-2} to 10^{-10} Torr depending on the type of electron generation method employed by the energy

transducer 18. The proximal end of the energy transducer 18 is coupled to the flexible insertion device 12, which includes conductors that provide electrical and/or optical signals to the conducting cathode 30. In order for an electrical connection to be provided to the conducting anode 32 at the distal end of the energy transducer 18, an outer conductive layer 36 is provided on the outer surface of the insulating tube 34, which connects with a conductor provided in the flexible insertion device 12.

In operation, electrons are generated at the conducting cathode 30 upon the application of an energy pulse received from the energy source 14 while the conducting anode 32 is held at ground potential. During electron generation or immediately thereafter, a negative high-voltage pulse is introduced to the conducting cathode 30. The magnitude of the pulse may vary between -10 kV and -100 kV and have a duration between 1 nsec to 200 nsec. The electrons generated at the conducting cathode 30 are accelerated across the tube 34 by the application of the voltage pulse to the conducting cathode 30, until they are decelerated upon impacting by the conducting anode 32, which results in the generation of the required x-rays. The conducting anode 32 and the conducting cathode 30 are held at ground potential once the pulse applied to the conducting cathode 32 terminates. The process is then repeated until the desired results are obtained. The pulse rate for the pulses applied to the conducting cathode 30 may vary from 1 Hz to 1000 Hz.

Referring now to Fig. 3, a preferred embodiment of the energy transducer 18 in a linear reverse configuration is disclosed. The energy transducer 18 includes a conducting anode 32 and a conducting cathode 30 respectively located at a proximal end and a distal end of a transducer body, which in the illustrated embodiment is an electrically insulating tube 34.

The conducting cathode 30 optionally includes a cathode tip 37, depending on the type of energy transducer that is being used. The sealed tube 34 is preferably 3-10 mm in length and 0.5-1.7 mm in diameter, and holds a vacuum within a cavity 38 that typically varies 10^{-2} to 10^{-10} Torr depending on the type of electron generation method employed by the energy transducer 18. The proximal end of the energy transducer 18 is coupled to the flexible insertion device 12, which includes conductors that provide electrical and/or optical signals to the conducting anode 32. In order for an electrical connection to be provided to the conducting cathode 30 at the distal end of the energy transducer 18, an outer conductive layer 36 is provided on the outer surface of the insulating tube 34, which connects with a conductor provided in the flexible insertion device 12.

In operation, electrons are generated at the conducting cathode 30 upon the application of an energy pulse to the conducting anode 32 while the conducting cathode 30 is held at ground potential. During electron generation or immediately thereafter, a positive high-voltage pulse is introduced to the conducting anode 32. The magnitude of the pulse may vary between -10 kV and -100 kV and it has a duration between 1 nsec to 200 nsec. The electrons generated at the conducting cathode 30 are accelerated across the tube 34 by the application of the voltage pulse to the conducting anode 32, until they are decelerated upon impacting the conducting anode 32, which results in the generation of the required x-rays. The conducting anode 32 and the conducting cathode 30 are held at ground potential once the pulse applied to the conducting cathode 32 terminates. The process is then repeated until the desired results are obtained. The pulse rate for the pulses applied to the conducting cathode 30 may vary from 1 Hz to 1000 Hz.

In order to maintain the required vacuum level, getter material may be introduced through one or more components to the energy transducer 18 prior to final assembly thereof, regardless of the type or configuration of said transducer. While the cathode 30 and anode 32 are preferably manufactured from tungsten, tantalum, copper-tungsten alloy, stainless steel or combination thereof, the getter material is preferably made of titanium, zirconium, tantalum, iron, vanadium, aluminum or an alloy composed of two or more of these materials. Suitable types of getters for this application are, for example, non-evaporable sintered porous type getters such as ST701, ST702 and ST707 getter alloys, manufactured by SAES-getters, Milan, Italy.

Figures 4-7 illustrate several preferred embodiments to be used for introduction of getter material to the energy transducer 18. These embodiments utilize in various configurations a conducting trunk on which the anode of said transducer is installed to apply the getter material to said transducer. Conducting trunks are originally used within energy transducers to enable a relatively simple method for joining the electrode components to an insulating tube in a manner that will provide a vacuum-tight seal to the x-ray generation mechanism. The above-mentioned figures describe the preferred embodiments with respect to a generic vacuum-tight structure that provides the mechanical housing for the x-ray production mechanism. However, those skilled in the art will readily recognize various modifications and changes, which may be made to the illustrated embodiments, such as using other conducting and insulating materials for the construction of the vacuum-tight structure and other modes of joining the conducting trunks to an insulating tube, without departing from the spirit and scope of the present invention.

Fig. 4 describes a preferred embodiment of the energy transducer 18 wherein the conducting cathode 30 and the conducting anode 32 are disposed within a vacuum-tight structure 40. The conducting cathode 30, preferably in a form of a cap, is installed at the distal end of a cathode tube 42, made from a low expansion glass sealing metal or metal alloy. For example, the cathode tube 42 is preferably made from KOVAR, an alloy including approximately 53% iron, 30% nickel and 17 % cobalt, manufactured by Goodfellow Cambridge, Ltd. of the United Kingdom. The conducting anode 32, preferably in a shape of a cap, is installed at the edge of an anode tube 46, also preferably made from KOVAR. The attachment of the cathode 30 to the cathode tube 42 and of the anode 32 to the anode tube 46 can be achieved by several methods, including mechanical pressure, welding, soldering or brazing. A glass bead (not shown), which is coupled to the cathode tube 42 in a vacuum-tight joint (by baking in reducing atmosphere, for example), is joined to one edge of a glass tube 44, creating a vacuum-tight seal between the cathode tube 42 and the glass tube 44, such that a predetermined length of the cathode tube 42, which will be referred to as the cathode trunk 43, remains within the glass tube 44. Similarly, another glass bead (not shown) is joined to the other edge of the glass tube 44, creating a vacuum-tight seal between the anode tube 46 and the glass tube 44 thereby leaving a portion of the anode tube 46, which will be referred to as the anode trunk 47, within the glass tube 44. The anode trunk 47 contains evacuation openings 48, preferably with a diameter of about 10-500 microns, located at the distal edge of said trunk within the glass tube 44. Getter material 45 is introduced into the anode trunk 47 and is packed between the anode 32 and the openings 48. The structure 40 is evacuated by connecting a vacuum pump to anode tube 46. Gases inside the glass tube 44 are pumped out

through the openings 48 and the anode tube 46. The structure 40 is evacuated to an appropriate level of vacuum prior to sealing the distal edge of said tube using a pinch-off or any other desired method. The cathode and anode tubes 42, 46 preferably have an inner diameter of about 0.1-0.5 mm and an outer diameter of about 0.3-0.8 mm, or any other dimensions, which are in accordance with the demand of an operative x-ray emitter with a diameter of less than 1.7 mm. The distance between the cathode 30 and the anode 32 is preferably less than one mm. The anode trunk 47 length is preferably about 2-5 mm.

As the length of the energy transducer 18 is preferably less than 10 mm and the distance between the cathode 30 and the anode 32 is fixed, there is, apparently, a tradeoff between the predetermined lengths of the cathode trunk 43 holding the cathode 30 and anode trunk 47 holding the anode 32. However, the length of the cathode trunk 43 holding the cathode 30 is preferably made as short as possible and the length of the anode trunk 47 holding the andoe 32 is preferably as long as possible. Although a cathode trunk can also be used for the introduction of getter material, such embodiments might be much more complicated, due to the incorporation of electrical and/or optical conductors within the cathode trunk, and do not offer the advantages that using an anode trunk for this purpose offers, as will be discussed in detail herein.

Stray gas molecules, which are either created during the peration of the energy transducer or diffuse into said vacuum-tight structure due to out gassing or leaks, are absorbed by the getter material 45 through the openings 48. In another preferred embodiment (not shown) the anode trunk can include a plurality of openings, scattered over the trunk surface, as long as the getter material particles' diameter is greater than the openings

diameter. As the getter material 45 is confined within the anode tube, it need not have the mechanical strength to support the anode structure. Therefore, it can be applied in the form of a powder, making the manufacturing process of the getter simpler and less expensive. The getter material 45 absorbs some of the thermal energy, which is created during x-ray production at the anode 32. Such thermal energy increases the temperature of the getter, thus causing its reactivation during operation by a mechanism involving the diffusion of sorbed gases into the bulk of the getter material and the exposure of a fresh, ultra-clean surface of getter.

Fig. 5 describes another preferred embodiment of the energy transducer 18 wherein the conducting cathode 30 and the conducting anode 32 are disposed within a vacuum-tight structure 40. The conducting cathode 30, which preferably has the form of a cap, is installed at the distal end of a cathode tube 52. The attachment of the cathode 30 to the cathode tube 52 can be achieved by several methods, including mechanical pressure, welding, soldering or brazing. A glass bead (not shown), which is coupled to the cathode tube 52 in a vacuum-tight joint, is joined to one edge of a glass tube 44, creating a vacuum- tight seal between the cathode tube 52 and the glass tube 44 and leaving a predetermined length of a cathode trunk 53 inside the glass tube 44. The conducting anode 32, is composed of a cap 32a, a neck 32b and a base 32c. The base 32c is installed at the edge of an anode tube 56 by one of various methods including mechanical pressure, welding, soldering or brazing. The neck 32b is embedded in getter material 55, which has the same external diameter of anode trunk 56. Another glass bead (not shown) that is coupled to the anode tube 56 is joined to the other edge of the glass tube 44, creating a vacuum-tight seal between the anode tube 56 and the

glass tube 44 and leaving a predetermined length of an anode trunk 57 inside the glass tube 44. The anode trunk 57 contains evacuation openings 58, preferably with a diameter of about 10-500 microns, located at the distal edge of said trunk within the glass tube 44. Stray gases are pumped out through the openings 48 and the structure 40 is evacuated to an appropriate level of vacuum prior to sealing the distal edge of anode tube 56 using, for example, a pinch-off method or any other desired method. The cathode and anode tubes 52, 56 preferably have an inner diameter of about 0.1-0.5 mm and an outer diameter of about 0.3-0.8 mm or any other dimensions, which are in accordance with the demand of an operative x-ray emitter with a diameter of less than 1.7 mm. The distance between the cathode 30 and the anode 32 is preferably less than one mm. The anode trunk 57 length is preferably about 2-5 mm.

Stray gas molecules, which are either created during the operation of the energy transducer or diffuse into said vacuum-tight structure due to out gassing or leaks, are absorbed by the getter material 55. The structure of conducting anode 32 provides mechanical strength to the getter, thus removing strength requirements from the getter. The getter volume constitutes about 30% of the total volume inside the vacuum-tight structure 40. This contributes to high volumetric efficiency, taking into account that the getter structure does not increase the voltage instability of the miniature energy transducer 18. The heat produced at the anode cap 32a is conducted along the anode neck 32b to the anode base 32c, thus providing a homogeneous heating of the porous getter material 55 from the inside. Such heating induces reactivation of the getter material during operation by a mechanism involving the diffusion of sorbed gases into the bulk of the getter material and the exposure of a fresh, ultra-clean surface of getter

Fig. 6 describes another preferred embodiment of the energy transducer 18 wherein the conducting cathode 30 and the conducting anode 32 are disposed within a vacuum-tight Structure 40. The conducting cathode 30, preferably in a form of a cap, is installed at the distal end of a cathode tube 62. A glass bead (not shown), which is coupled to the cathode tube 62 in a vacuum-tight joint, is joined to one edge of a glass tube 44, creating a vacuum-tight seal between the cathode tube 62 and the glass tube 44 and leaving a predetermined length of a cathode trunk 63 inside the glass tube 44. The conducting anode 32, preferably in a shape of a cap, is installed at the distal end of a Getter trunk 65. The proximal end of Getter trunk 65 is installed at the distal end of an anode tube 66. Another glass bead (not shown) that is coupled to the anode tube 66, is joined to the other edge of the glass tube 44, creating a vacuum- tight seal between the anode tube 66 and the glass tube 44 and leaving a relatively short segment of an anode trunk 66a inside the vacuum-tight structure 40. The anode trunk 66a contains evacuation openings 68 that are used for evacuating stray gases, thus permitting the structure 40 to be evacuated to an appropriate level prior to sealing the distal end of anode tube 66 using a pinch-off method or any other desired sealing method.

Stray gas molecules, which are which are either created during the operation of the energy transducer or diffuse into said vacuum-tight structure due to out gassing or leaks, are absorbed by the getter material of the Getter trunk 65. Although the getter material used for this embodiment must provide sufficient mechanical strength to act as the Getter trunk 65, the volumetric efficiency of this embodiment is higher than those of the embodiments described in Fig. 4 and Fig.5. The getter material in this embodiment must also have the ability to conduct electrical current as electrical coupling is provided to conducting anode 32 through

Getter trunk 65. This characteristic is not essential with the embodiments described in Fig. 4 and Fig. 5, wherein electrical coupling is provided to conducting anode 32 directly through anode trunks 47 and 57, respectively. The Getter trunk 65 absorbs the thermal energy, which is created during x-ray production in the anode 32. Such thermal energy increases the temperature of the getter, thus causing its reactivation during operation by a mechanism involving the diffusion of sorbed gases into the bulk of the getter material and the exposure of a fresh, ultra-clean surface of getter

In some types of energy transducers a getter coating as illustrated in Fig.7 may be sufficient to maintain the required vacuum level. In the preferred embodiment illustrated in Fig.7 the conducting cathode 30 and the conducting anode 32 are disposed within a vacuum-tight structure 40. The conducting cathode 30, preferably in a form of a cap, is installed at the edge of a cathode tube 72. A glass bead (not shown), which is coupled to the cathode tube 72 in a vacuum-tight joint, is joined to one edge of a glass tube 44, creating a vacuum-tight seal between the cathode tube 72 and the glass tube 44 and leaving a predetermined length of a cathode trunk 73 inside the glass tube 44. The conducting anode 32, preferably in a shape of a cap, is installed at the edge of an anode tube 76. Another glass bead (not shown) that is coupled to the anode tube 76 is joined to the other edge of the glass tube 44, creating a vacuum-tight seal between the anode tube 76 and the glass tube 44 and leaving a predetermined length of an anode trunk 77 inside the glass tube 44. The anode trunk 77 contains evacuation openings 78 that are used for evacuating stray gases, thus permitting the vacuum-tight structure 40 to be evacuated to an appropriate level of vacuum prior to sealing the distal end of anode tube 76 using a pinch-off method or any other desired sealing method.

A getter coating 75 is deposited on the anode trunk 77 and/or cathode trunk 73, preferably by sputtering, physical evaporation or related methods. The coated parts are assembled together prior to the evacuation of the structure 40 to an appropriate level of vacuum. Stray gas molecules, which are either created during the operation of the energy transducer or diffuse into said vacuum-tight structure due to out gassing or leaks, are absorbed by the getter coating.

All the different embodiments used for the introduction of getter material into the energy transducer 18 need a heating step in order to activate the getter. The getter material is activated by heating the energy transducer 18 to a temperature that is preferably 300°C or more. This heating step may be a by-product of the routine bake-out processes necessary to out-gas the x-ray emitting energy transducer 18 before it is sealed off or can be performed as a separate manufacturing step. During the above mentioned heat activation process, the native oxide on the getter material surface is decomposed and oxygen diffuses into the bulk of the getter material, exposing a fresh, ultra-clean surface of getter material.

The resultant surface of the getter material has beneficial absorption characteristics, and can typically absorb gases, for example H_2, CO, and H_2O vapor. Typical surface absorption capacity for the above mentioned getter material is approximately 10^{15} atoms per cm^2. Thus, if the effective gettering surface is at least 5 mm^2 in each of the above mentioned embodiments, then 1.5 x 10^{-6} Torr-liter/sec can be absorbed. Taking out-gassing characteristics and in-operation gas-disorption rates into account and maintaining a leak rate of about 10^{-14} Torr-liter/sec make this absorption rate more than sufficient for keeping the required vacuum level for more than one year on the shelf and throughout operation. Re-

heating the sealed energy transducer 18 may further extend the shelf life by re-activating the getter matrial. The main advantage of providing getter material in the various modes described in Figures 4-7 is enhanced preservation of the vacuum level to provide x-ray energy transducers with a prolonged operational period and shelf life.

The introduction of the getter material, as described in the above-mentioned embodiments, is applicable to different types of energy transducers wherein different cathode structures or anode structures in several possible configurations are employed.

Fig.8, for example, illustrates a conducting cathode 30 that includes a conducting cathode shell 80 that forms a cavity 82. The conducting cathode shell 80 is provided with an electron escape nozzle 84 having a diameter in the range of about 10 to 200 microns. In this embodiment, laser light is introduced into a proximal end of the conducting cathode shell 80 via an optical fiber 86 provided within the flexible insertion device 12. The conducting cathode shell 80 is also electrically coupled to an electrical conductor 87 provided around the optical fiber 86 within the flexible insertion device 12. In this illustration, for simplicity, the conductor in the flexible insertion device 12 coupled to the conducting anode 32 is not shown. The laser light may optionally be focused by a lens 88 provided at the end of the optical fiber 86. The lens 88 may constitute a separate element or the end of the optical fiber 86 can be shaped in the form of a lens. The laser light strikes the distal end of the cavity 82 inside the conducting cathode shell 80. When high laser intensities, greater than $100MW/cm^2$, are introduced into the conducting cathode shell 80, a plasma containing electrons, ions and neutral atoms is formed, for example, by the ablation of the material that composes the conducting cathode shell 80 at the point of impact. In a preferred embodiment,

a laser light pulse is applied that has a duration of 1-20 nsec and an intensity of 10^9 - 10^{11} W/cm^2.

The electrons possess a thermal velocity that is approximately 50 times greater than that of the ions. As a result, a significant percentage of the electrons escape through the electron escape nozzle 84 before a considerable number of ions escape. The electrons that escape are replenished from the cavity walls of the conducting cathode shell 80 due to thermionic emission and/or photo emission induced by the high temperature of the plasma. The small diameter of the electron escape nozzle 84 effectively renders the cavity 82 a Faraday cage, thereby isolating electrons and ions in cavity 82 from fields formed outside the cavity 82. The electron escape nozzle 84 may also prevent, or at least decrease, the coating of the insulating tube 34 by electrically conducting material.

The electron escape nozzle 84, however, can also be eliminated as illustrated in the embodiment shown in Fig. 9. In this embodiment, a conducting cathode shell 90 is also provided that forms a cavity 92. A laser pulse is introduced via an optical fiber 94 provided in the flexible insertion device 12, and the conducting cathode shell 90 is coupled to a conductor 95 provided on the optical fiber 94. Again a lens 96 may optionally be used to focus the laser energy to heat the distal outer surface 98 of the conducting cathode shell 90. The temperature of the outer surface 98 is raised to a level comparable with the cathode material's work function and, as a result, electrons are emitted from the distal end of the conducting cathode shell 90. The laser light pulse is applied in the same manner discussed above.

Fig. 10 shows another preferred embodiment of an energy transducer wherein laser induced plasma is used to generate electrons. The energy transducer 18 utilizes a distal illumination plasma cathode 100, located at the distal end of said transducer, and an anode 102, located at the proximal end of said transducer. The cathode 100 and the anode 102 are disposed within an insulating shell 104 wherein a vacuum of about 10^{-4} Torr is maintained. The spacing between the cathode 100 and the anode 102 is less than 5 mm and preferably, less than 2 mm. The anode 102 is an annular structure that houses an optical fiber 106 with a core diameter of about 0.5 mm or less. The proximal end of the optical fiber 106 is connected to a light source (not shown), which may be a laser. The distal end of the optical fiber 106 is disposed in the annular anode structure. The laser light may optionally be focused by a lens 108 provided at the end of the optical fiber 106. The lens 108 may constitute a separate element or the end of the optical fiber 106 can be shaped in the form of a lens. The anode 102 is electrically coupled to an electrical conductor provided within the flexible insertion device 12. In order for an electrical connection to be provided to the cathode 100 at the distal end of the energy transducer 18, an outer conductive layer 107 is provided on the outer surface of the insulating tube 104, which connects with a conductor provided in the flexible insertion device 12. The electrical conductors within the flexible insertion device 12 preferably comprise structural elements of a coaxial cable. The optical fiber 106 is preferably incorporated in the flexible insertion device 12. For example, the optical fiber 106 can be part of the core of a coaxial cable.

In operation, a light pulse (illustrated by arrows 109) travels through the optical fiber 106 and strikes the proximal end of the cathode 100. If the intensity of the light pulse

exceeds approximately 10^9 Watt/cm^2, micro-plasma is created. The micro-plasma has a high density of approximately 10^{18} electrons per cm^3. Therefore, the electrons, ions and neutral atoms temperatures equilibrate in a matter of picoseconds. The electrons, by virtue of their higher thermal velocities are drawn towards the outer shell of the micro-plasma. A high voltage pulse, having a peak voltage pulse of 10-100 kV, is timed to begin shortly after the micro-plasma is created. The electrons are detached from the micro-plasma by the electrical field created in the gap between the cathode 100 and the anode 102 and strike the anode 102 to produce x-rays. Upon gap closure by the micro-plasma the high voltage pulse is terminated, and after a rest period the cycle starts all over again.

Referring to Fig.11, another preferred embodiment is described, wherein the flexible insertion device 12 is provided with an internal optical fiber 116, optionally having a lens 119 located at its distal end. An x-ray transmissive insulating tube or shell 114 is provided around the end of the optical fiber 116 to form a cavity 118 containing a conducting anode 112 and an emission element 110 which serves as the cathode. The conducting anode 112 is coupled to a conductor 111 provided around the optical fiber 116. The emission element 110, for example, is a photo-emission cathode or thermionic emission surface, and is grounded by a metallic layer 113 that contacts with a conductor in the flexible insertion device 12. The insulating shell 114 is sealed and holds a vacuum that typically varies from 10^{-6} Torr, when a photo cathode is used for the emission element 110, to 10^{-4} Torr, when a thermionic emission surface is used for the emission element 110.

In this embodiment, electrons are emitted by the photoelectric effect when a photocathode is used as the emission element 110, and electrons are emitted by the

thermionic emission effect when a thermionic emission surface is used as the emission element 110. Laser light is focused on the emission element 110 by the lens 119 (although the lens 119 can be omitted if desired), wherein the electron generation process preferably takes less than 10 nsec. The same pulse duration and amplitude are utilized as with the previously described embodiments. Immediately thereafter, or even during the process of electron generation, a positive high-voltage pulse is introduced to the conducting anode 112. The magnitude of the pulse may vary between +10kV and +100kV. The electrons created at the emission element 110 are accelerated across the cavity 118 by the pulse and are decelerated upon impacting the conducting anode 112 thereby causing the required x-rays to be emitted.

If the emission element 110 is a photo cathode, it is preferably made of a highly efficient photo emissive material, with an optimal wave length efficiency response. Conventional materials of the type suitable for this purpose include, but are not limited to, metals, such as Au, Mg, Cu, semiconductors, like gallium arsenic, and compounds, such as Cs3 Sb, Cs2 Te, and AgOCs. It will be appreciated by those skilled in the art that such photo emissive materials vary in efficiency and in operating conditions parameters, such as vacuum requirements and the light wave length of the light source, which are well known.

If the emission element 110 is a thermionic emission surface, it is preferably made of tungsten, tantalum or other refractory metals having high melting and sublimation points. Electron currents greater that 500 Amps/cm^2 have been measured experimentally in tantulum and tungsten. Additionally, the ratio between the work function and the melting temperature is low for these materials, which renders them good thermionic emitters.

Both Fig. 10 and Fig. 11 describe preferred embodiments of miniature energy transducers wherein optical fibers 106 and 116 are inserted into the anode structures 102 and 112 respectively. These specific configurations impose a difficulty regarding the placement of getter material within the energy transducer 18 using the embodiments illustrated in Fig.4-6. Some of the space within anode trunk 47 or within the Getter trunk 65 might be occupied by optical fibers. Additionally, a vacuum-tight seal between the optical fiber 106,116 and the anode tube or trunk is needed, making the assembly of the energy transducer more complicated. Thus, one preferred mode to provide getter material to the above mentioned energy transducers is by a getter coating, as described in Fig. 7. Another option is utilizing the cathode structure in the same modes the anode structure is used for this purpose, as described with respect to Fig.4-6. Fig. 10 shows, for example, getter material 105 confined within a cathode trunk 101, the getter 105 being introduced into miniature energy transducer 18 in an equivalent manner to that used with anode trunk 47 in Fig. 4.

It is also possible to provide a miniature energy transducer that emits x-rays without the use of a light source as required in the previously described embodiments. Fig.12, for example, illustrates a spark induced plasma cathode 120 that can be utilized in the miniature energy transducer 18 with a linear configuration as illustrated in Fig. 2. In this embodiment, the flexible insertion device 12 has internal electrical conductors 122 and 124 that are separated by an insulator 126, thereby creating a spark gap 128 at a distal end of the flexible insertion device 12. A conducting cathode 120 is provided that, like the embodiment of Fig. 8, includes an electron escape nozzle 123 having a diameter of approximately 10 to 200 microns. In this embodiment, high voltage pulse, of a duration between 1-200 nsec and a

magnitude of 5-50 kV, is applied across the spark gap 128 via the internal electrical conductors 122, 124 of the flexible insertion device 12. The electrical breakdown creates hot plasma, consisting of electrons, ions and neutral atoms, that is formed within a cavity 127 formed by the cathode 120. As in the previous embodiment, the electrons possess a thermal velocity that is 50 times greater than that of the ions. As a result, a significant percentage of the electrons escape through the electron escape nozzle 123 before a considerable number of ions escape. The electrons that escape are replenished from the cavity 127 walls by thermionic emission and photo-emission induced by the plasma.

Furthermore, the spark induced plasma cathode 120 can be utilized in the miniature energy transducer 18 with a linear reverse configuration as illustrated in Fig. 3. In this form of embodiment, internal electrical conductors, separated by an insulator and thereby creating a spark gap are provided at the distal end of the energy transducer 18. A conducting cathode 120 is provided that, like the embodiment of Fig. 4, includes an electron escape nozzle having a diameter of approximately 10 to 200 microns. A high voltage pulse, of a duration between 1-200 nsec and a magnitude of 5-50 kV, is applied across the spark gap via the internal electrical conductors provided at the distal end of the energy transducer 18. Other than this, the process of generating the plasma and the acceleration of the electrons towards the anode is the same as described for the embodiment in Fig 12. The linear reverse configuration appears to be more complicated than the linear configuration. It requires the provision of the distal end of the energy transducer 18 with both internal electrical conductors for spark gap creation and with an outer electrical conductor in order to electrically connects the conducting cathode 120 to a

high voltage source. However, the linear reverse configuration has some advantages that do not exist if the linear configuration is used. Holding the cathode and the outer electrical conductor at the same potential yields inward pointing electrical field lines, which results in enhanced high-voltage survivability of the x-ray emitter. Particularly, secondary electron emission from the walls of the insulating tube is reduced.

Other preferred embodiments of miniature energy transducers that emit x-rays without the use of a light source include explosive electron emission (EEE) plasma cathode energy transducers. An EEE plasma cathode can be utilized in the miniature energy transducer 18 with a linear reverse configuration illustrated in Fig. 3. The bulk of the cathode 30 consists of a thermal/electrical conductor on which an EEE plasma cathode tip 37 is located. The EEE cathode tip 37 may be a liquid metal cathode (LMC) or it may be composed of other materials, such as carbon fibers, which are not necessarily metallic. For example, the EEE plasma cathode can include a single or a plurality of fibers, made out of a material characterized by a high electrical resistance and fiber diameter ranging from a few up to about 30 microns, such as tungsten, molybdenum or carbon fibers.

Fig. 13 illustrates a preferred embodiment of a EEE plasma cathode LMC tip 132 having an outer capillary tip 134 and a liquid metal 136 interior. The outer capillary tip 134 is composed of a solid material, preferably tungsten, kovar, stainless steel, glass or ceramic material and the liquid metal 136 is composed of preferably gallium or gallium-indium. The capillary tip 134 is preferably cone shaped with a small opening at its distal end. The liquid metal 136 fills the interior of the cone, and a small amount of the liquid metal 136 protrudes out of the opening forming a location of explosive emission 138. In

operation, the cathode 30 is held at ground potential while a surface generating pulse is applied to the anode 32. The application of the surface generating pulse to the anode 32 generates an electric field gradient that causes the liquid metal protrusion 138 to assume the shape of a sharp tip, the dimensions of which are approximately 10 microns. Once the required surface shape is created, a second pulse is applied to the anode 32. This second pulse causes rapid joule heating of the liquid metal, which in turn causes the liquid metal to be ablated and ionized into an explosively emitted plasma and also causes the electrons within the plasma to be accelerated towards the anode 32, causing the required x-rays to be emitted.

Fig. 14 illustrates a preferred embodiment of an EEE plasma cathode with a carbon fiber tip 140. The carbon fiber EEE plasma cathode tip 140 includes either a single carbon fiber or a central bundle of carbon fibers 142 surrounded by a capillary tip 144. The central bundle of carbon fibers 142 is preferably composed of carbon fibers of 1-10 micron in diameter, optionally with CsI, and the outer capillary tip 144 is composed of a solid material, preferably tungsten, KOVAR, glass or a ceramic material. A small length, typically less than 1 mm and up to a few mm, of the carbon fibers 142 protrude beyond the capillary tip 144 forming a location of explosive emission 146. In contrast to the EEE plasma cathode LMC tip 134 illustrated in Fig. 13, the carbon fiber EEE plasma cathode tip 140 generates plasma within nanoseconds of the application of a voltage pulse to the anode 32. Carbon fibers start to emit electrons at average electric fields less than 10 kV/cm. This means that at accelerating field of 30 kV/cm, for example, at a distance of one mm between the anode 32 and the cathode 30, carbon fibers start to operate with a

delay less than 100 psec with respect to the beginning of the accelerating pulse. Thus, a single pulse, 10-100 kV magnitude and duration of 4-100 nsec, can be used. The electrons within the plasma are accelerated towards the anode 32, causing the required x-rays to be emitted.

Furthermore, an EEE plasma cathode, either with a LMC tip or with a carbon fiber tip, can be utilized in the miniature energy transducer 18 with a linear configuration illustrated in Fig. 2. The operation of the energy transducer 18 is such that the anode 32 is held at ground potential while a surface generating pulse is applied to the cathode 30. Additionally, the plasma generation and electron acceleration pulse is applied to the cathode 32 and is negative.

As can be seen by the illustrated examples, the miniature energy transducer 18 may consist of a variety of different types, each of which can utilize a pulsed energy source to accelerate the electrons, wherein getter material can be provided through one or more structural elements, as described in Fig 4-7.

The invention has been described with reference to certain preferred embodiments thereof. It will be understood, however, that modification and variations are possible within the scope of the appended claims. For example, the disclosed methods for getter material application may also be utilized for conventional full sized x-ray tubes. In addition, other metals or metal alloys may be utilized as getter materials.

WHAT IS CLAIMED IS:

1. An x-ray emitting energy transducer comprising:

a transducer body;

a cathode provided at a first end of the transducer body;

an anode provided at a second end of the transducer body opposite the cathode; and

wherein the transducer body, cathode and anode form a cavity; and

wherein at least one of the cathode and the anode is mounted on a trunk

2. A transducer as claimed in claim 1, wherein a getter material is provided within the conductive trunk.

3. A transducer as claimed in claim 2, wherein the getter material is in powder form.

4. A transducer as claimed in claim 1, wherein at least one of the anode and the cathode comprises a cap, a neck and a base.

5. A transducer as claimed in claim 4, wherein the neck is embedded in a getter material.

6. A transducer as claimed in claim 1, wherein the trunk includes a getter trunk portion and an alloy trunk portion.

7. A transducer as claimed in claim 1, wherein a getter coating is provided on a surface of the trunk.

8. An energy transducer as claimed in claim 2, wherein the getter material comprises at least one of titanium, zirconium, tantalum, iron, vanadium and aluminum.

9. An energy transducer as claimed in claim 5, wherein the getter material comprises at least one of titanium, zirconium, tantalum, iron, vanadium and aluminum.

10. An energy transducer as claimed in claim 6, wherein the getter material comprises at least one of titanium, zirconium, tantalum, iron, vanadium and aluminum.

11. An energy transducer as claimed in claim 1, wherein the cathode comprises at least one of tungsten, tantalum, copper-tungsten alloy, iron-nickel-cobalt alloy and stainless steel.

12. An energy transducer as claimed in claim 1, wherein the anode comprises at least one of tungsten, tantalum, copper-tungsten alloy, iron-nickel-cobalt alloy and stainless steel.

13. An energy transducer as claimed in claim 1, wherein an outer diameter of the energy transducer is 1.7 mm or less.

14. An energy transducer as claimed in claim 1, wherein the transducer body has a length less than 10 mm.

15. An energy transducer as claimed in claim 1, wherein the trunk comprises a conductive material.

16. An energy transducer as claimed in claim 15, wherein the trunk comprises a conductive alloy.

17. An energy transducer as claimed in claim 15, wherein the alloy includes iron, nickel and cobalt.

18. An energy transducer as claimed in claim 1, wherein the transducer body comprises a glass tube.

19. An energy transducer as claimed in claim 18, wherein the trunk includes openings that permit evacuation of the cavity to a desired vacuum level.

20. An energy transducer as claimed in claim 19, wherein the openings have a diameter of 10 to 500 microns.

21. An energy transducer as claimed in claim 1, wherein the trunk has an inner diameter of about 0.1-0.5 mm and an outer diameter of about 0.3-0.8 mm.

22. An energy transducer as claimed in claim 1, wherein the distance between the anode and cathode is less than one millimeter.

23. An energy transducer as claimed in claim 1, wherein the trunk length is between 2-5 mm.

24. An energy transducer as claimed in claim 1, wherein the getter material comprises thirty percent of the volume within the cavity.

25. An x-ray radiation treatment system comprising:

a flexible insertion device;

a x-ray emitting transducer coupled to a distal end of the flexible insertion device, wherein the transducer includes a cathode provided at a first end of the transducer body, and an anode provided at a second end of the transducer body opposite the cathode, wherein the transducer body, cathode and anode form a cavity, and wherein at least one of the cathode and the anode is mounted on a said transducer coupled to a distal end of the flexible insertion device; and

an energy source coupled to a proximal end of the flexible insertion device.

26. The system as claimed in claim 25, further comprising a dosimetry system that measures the output of the x-ray emitting energy transducer.

27. The system as claimed in claim 26, further comprising a control unit for controlling the amount of energy supplied by the energy source to the energy transducer in response to the output measured by the dosimetry system.

28. A miniature x-ray transducer comprising:

an insulating transducer body;

an anode located at a first end of the body;

an explosive electron emission cathode located at a second end of the body opposite the anode;

wherein the insulating transducer body, the anode and the explosive electron emission cathode define a cavity.

29. A miniature x-ray transducer as claimed in claim 28, wherein the insulating transducer body has a length of less than 7 mm and a diameter less than 1.7 mm.

30. A miniature energy transducer as claimed in claim 28, wherein the explosive electron emission cathode comprises a liquid metal cathode.

31. A miniature energy transducer as claimed in claim 30, wherein the liquid metal explosive electron emission cathode is composed of an electrically conducting material selected from the group consisting of gallium-indium and gallium .

32. A miniature energy transducer as claimed in claim 28, wherein the explosive electron emission cathode includes at least one high resistance material fiber.

33. A miniature energy transducer as claimed in claim 32, wherein the high resistance material fiber has a diameter of 1-10 microns.

34. A miniature energy transducer as claimed in claim 32, wherein the high resistance material fiber is a carbon fiber.

35. A miniature energy transducer as claimed in claim 28, further comprising means for applying electrical signals to the miniature energy transducer

36. A miniature energy transducer as claimed in claim 35, wherein the means for applying electrical signals to the miniature x-ray transducer includes a coaxial cable structure.

37. A miniature energy transducer as claimed in claim 36, wherein the coaxial cable structure is coupled to the first end of the transducer body at which the anode is located.

38. A miniature energy transducer as claimed in claim 37, wherein the coaxial cable structure is coupled to the second end of the transducer body at which the cathode is located.

39. A miniature energy transducer as claimed in claim 38, further comprising a conductor that extends from the second end of the transducer body to the first end of the transducer body, wherein the conductor is connected to the cathode.

40. A miniature energy transducer as claimed in claim 39, further comprising a conductor that extends from the first end of the transducer body to the second end of the transducer body, wherein the conductor is connected to the anode.

41. A miniature energy transducer as claimed in claim 35, wherein the means for applying electrical signals includes high voltage power supply.

42. A miniature energy transducer as claimed in claim 28, further comprising a flexible insertion device having one end coupled to the miniature energy transducer and

another end coupled to an electrical signal source.

43. A miniature energy transducer as claimed in claim 42, wherein the flexible insertion device is coated with a bio-compatible material.

5

44. An x-ray transducer as claimed in claim 30, further comprising means for applying a surface generating voltage pulse to at least one of the anode and the cathode to generate an electric field gradient that causes the liquid metal cathode to assume the shape of a sharp tip.

10

45. An x-ray transducer as claimed in claim 28, further comprising means for applying voltage pulses to at least one of the cathode and the anode to cause generation of a plasma within the cavity and acceleration of electrons within the plasma toward the anode.

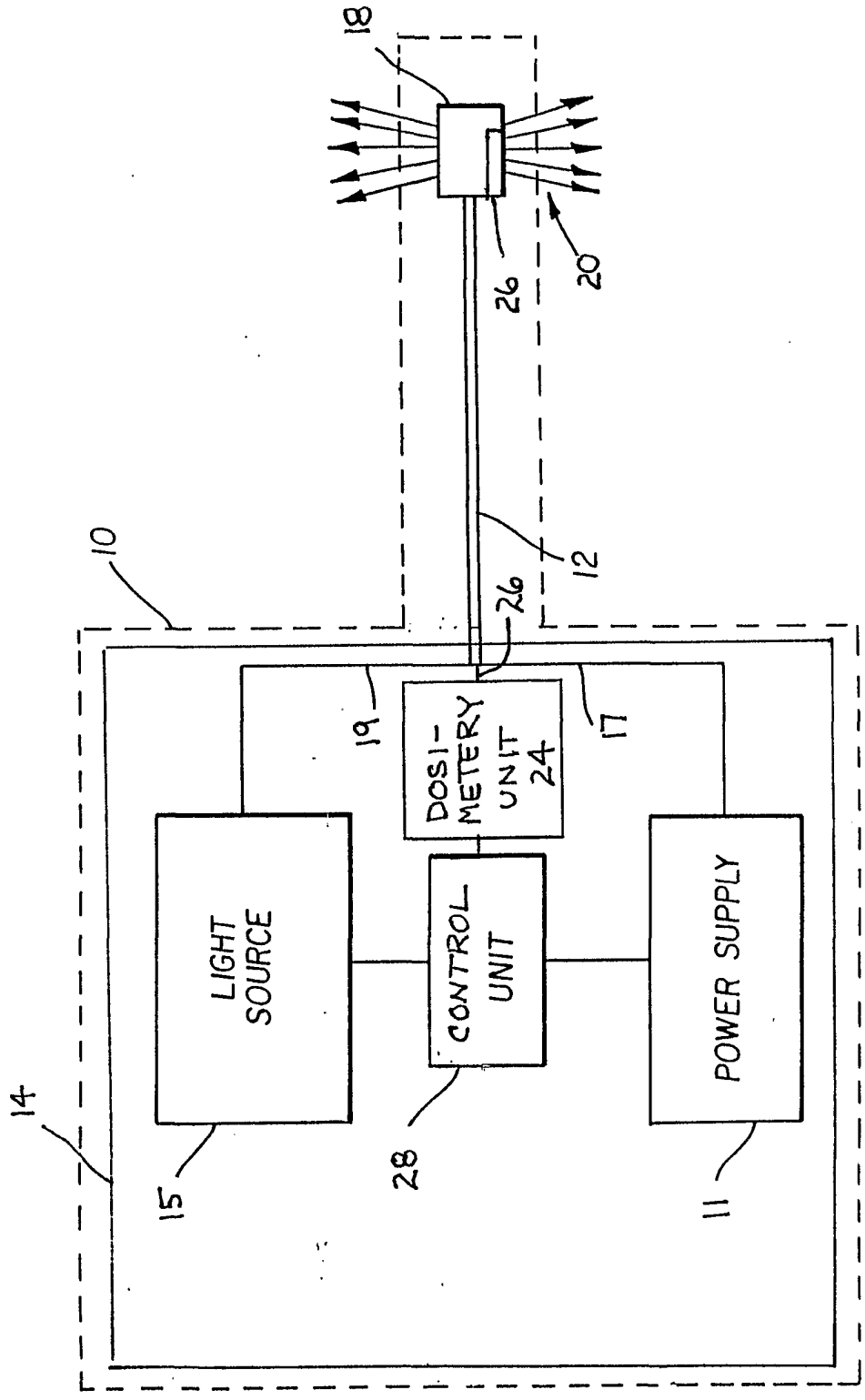

18
20
26
12
10
26
19
DOSI-METERY UNIT 24
17
LIGHT SOURCE
CONTROL UNIT
POWER SUPPLY
14
15
28
11

FIG. 1

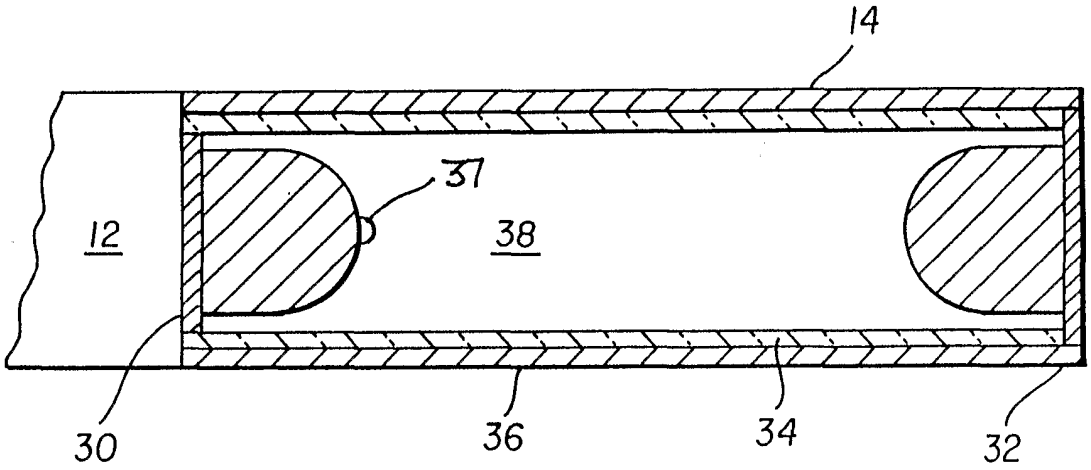


FIG. 2

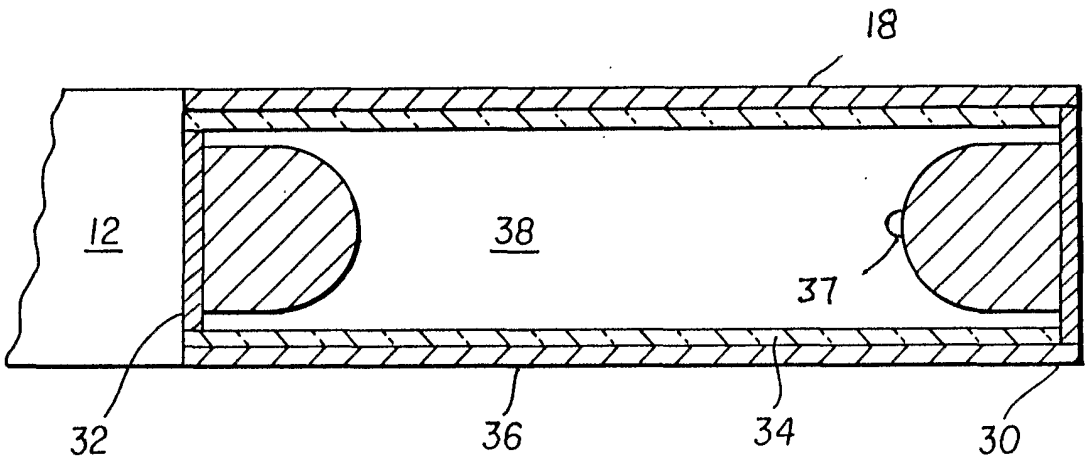


FIG. 3

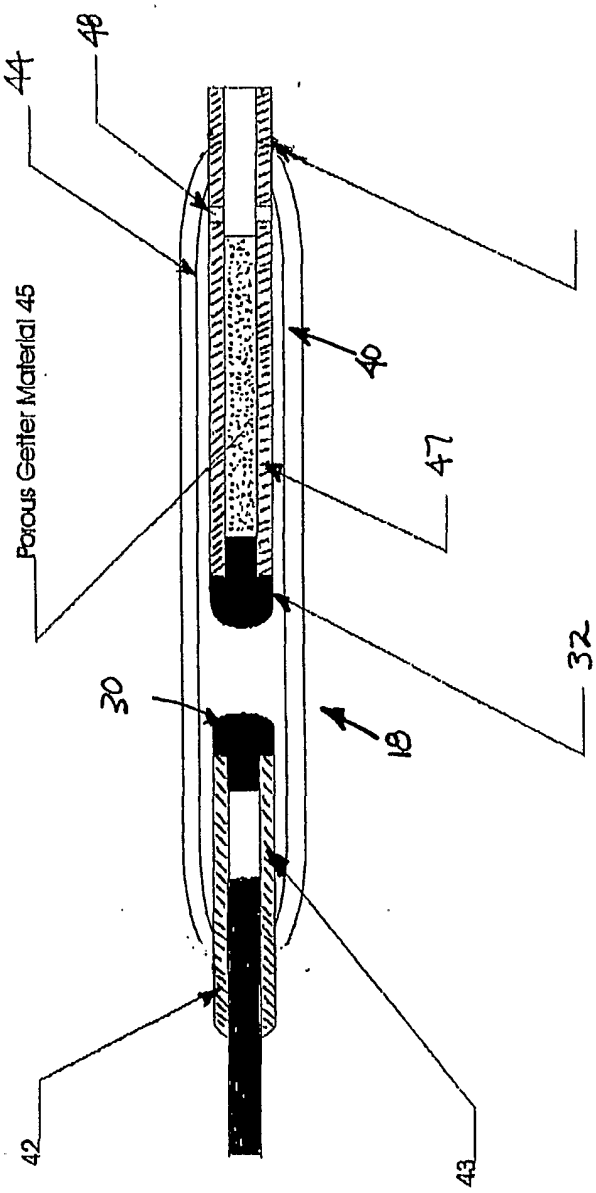


Fig. 4

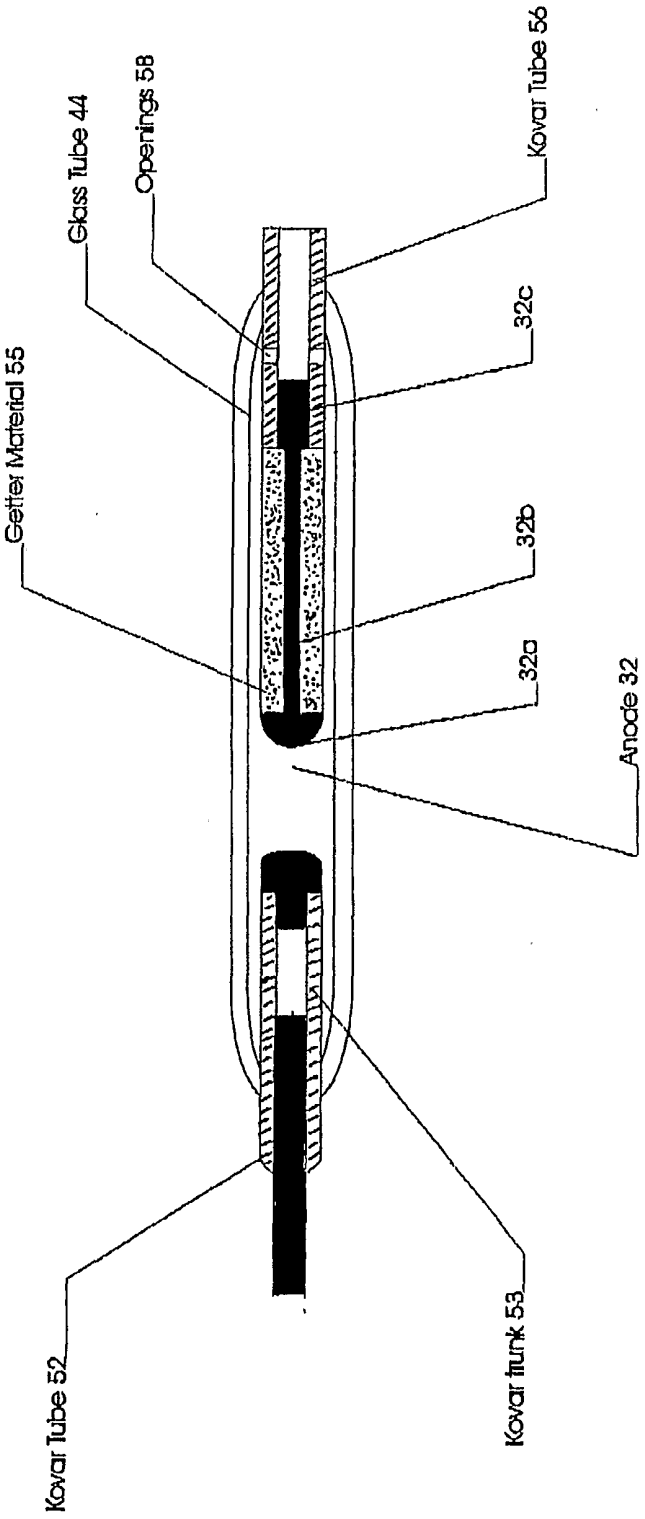

Kovar Tube 56
Openings 58
Glass Tube 44
32c
Getter Material 55
32b
32a
Anode 32
Kovar Tube 52
Kovar trunk 53

Fig. 5

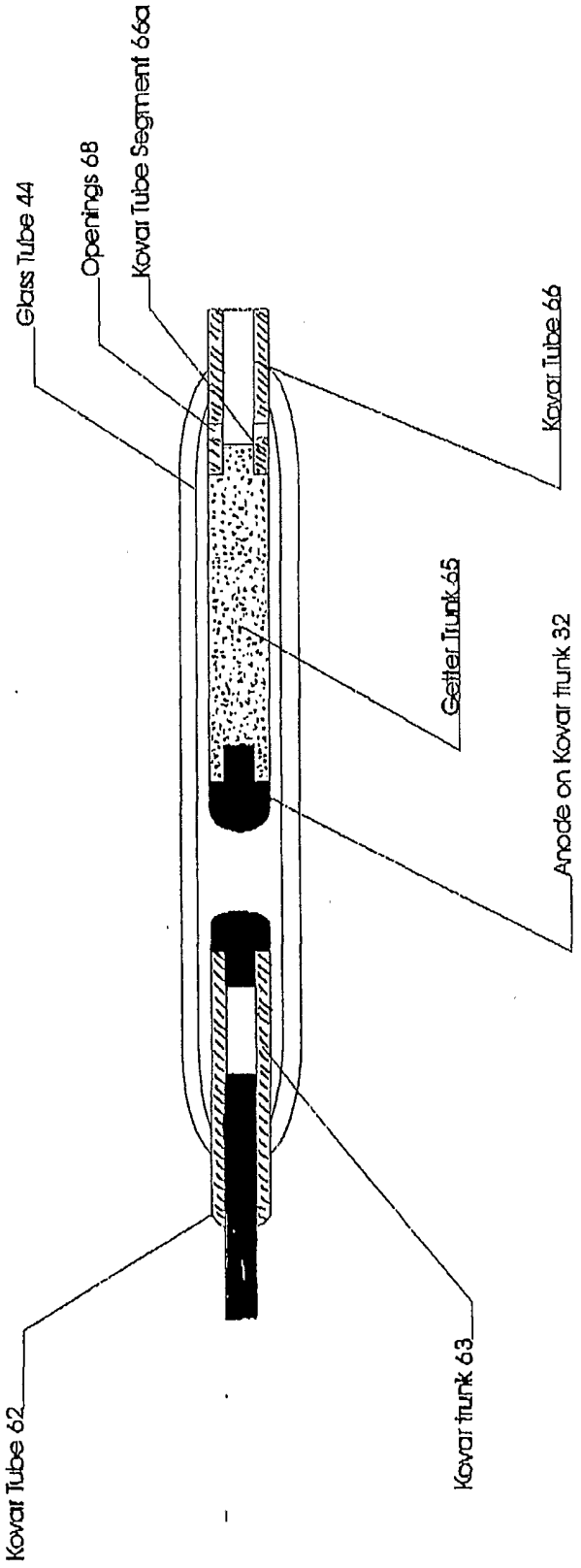

Glass Tube 44
Openings 68
Kovar Tube Segment 66a
Kovar Tube 66
Getter Trunk 65
Anode on Kovar trunk 32
Kovar Tube 62
Kovar trunk 63

Fig. 6

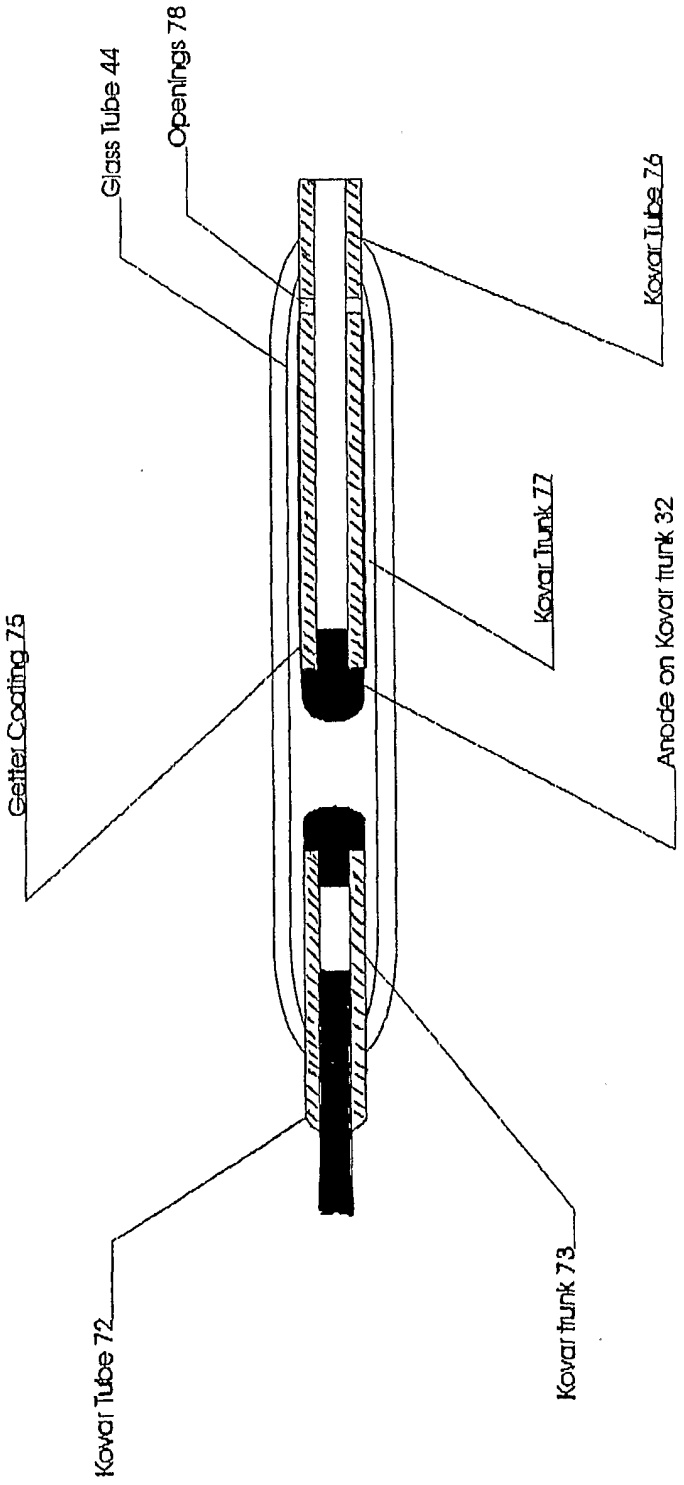

Glass Tube 44
Openings 78
Kovar Tube 76
Kovar Trunk 77
Anode on Kovar trunk 32
Getter Coating 75
Kovar Tube 72
Kovar trunk 73

Fig. 7

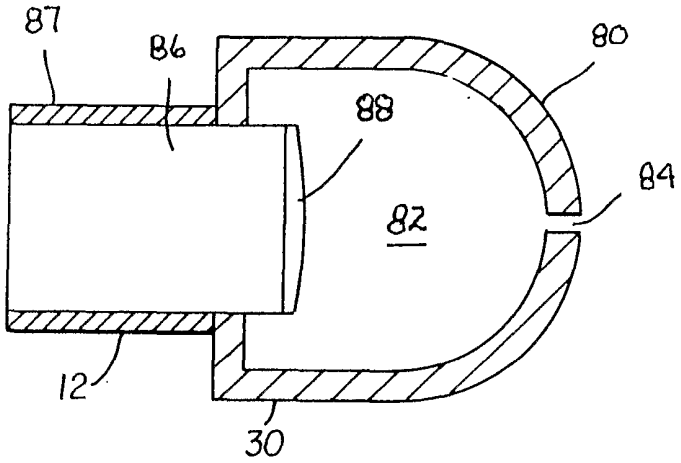


FIG. 8

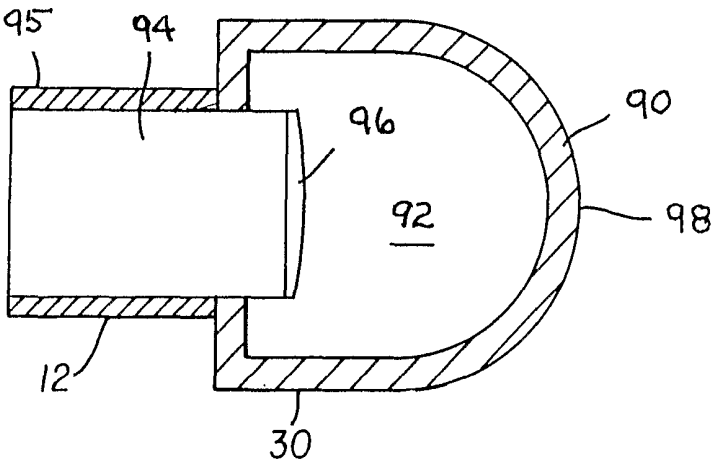


FIG. 9

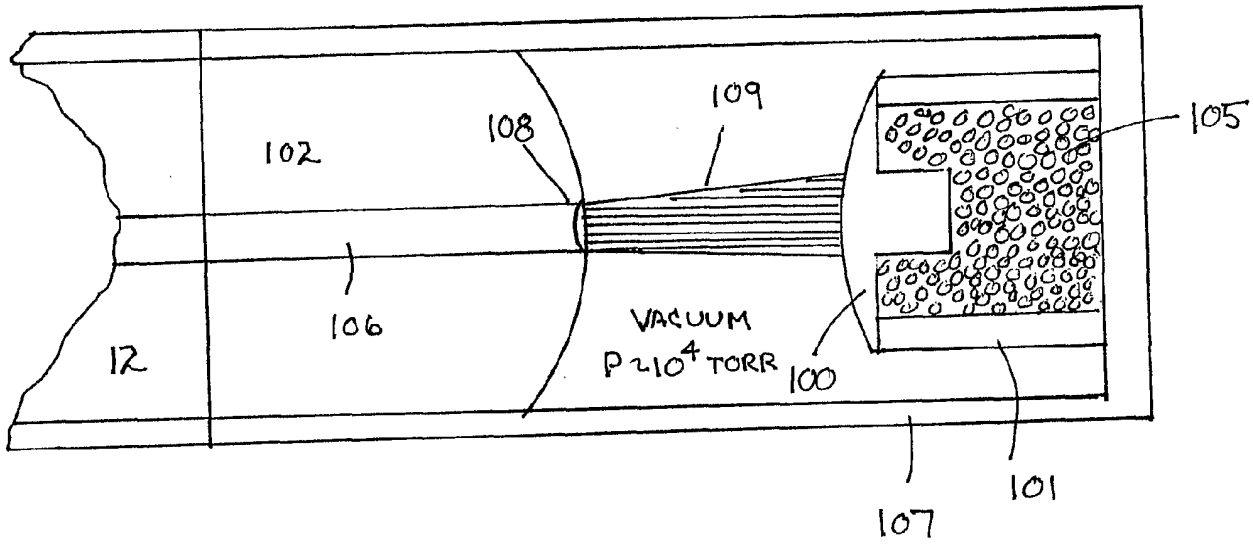


FIG 10

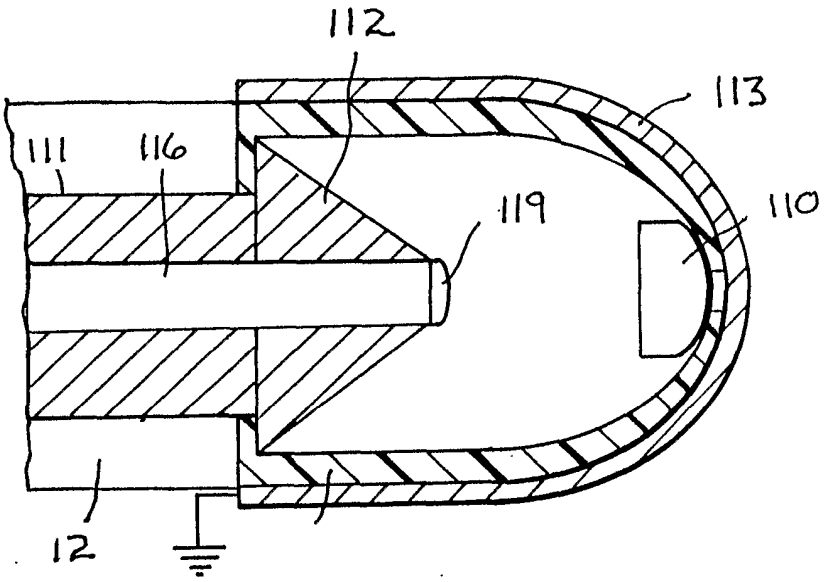


FIG. 11

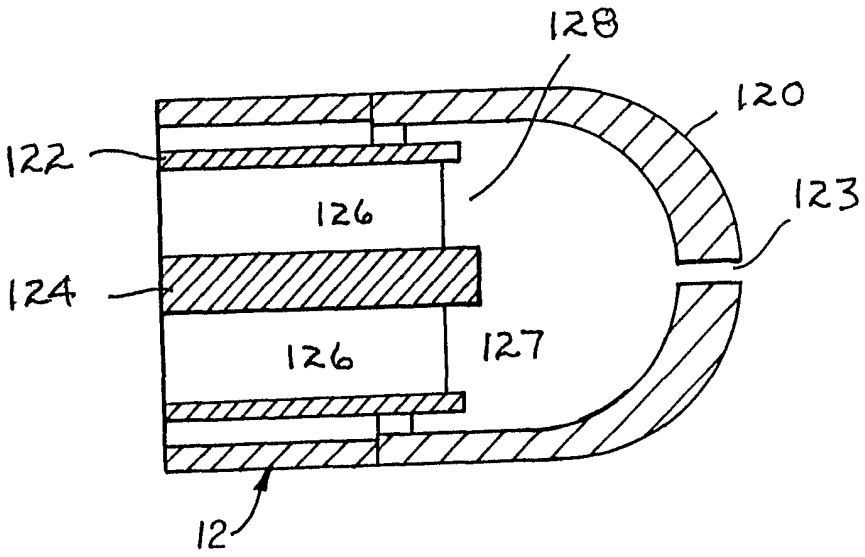


FIG. 12

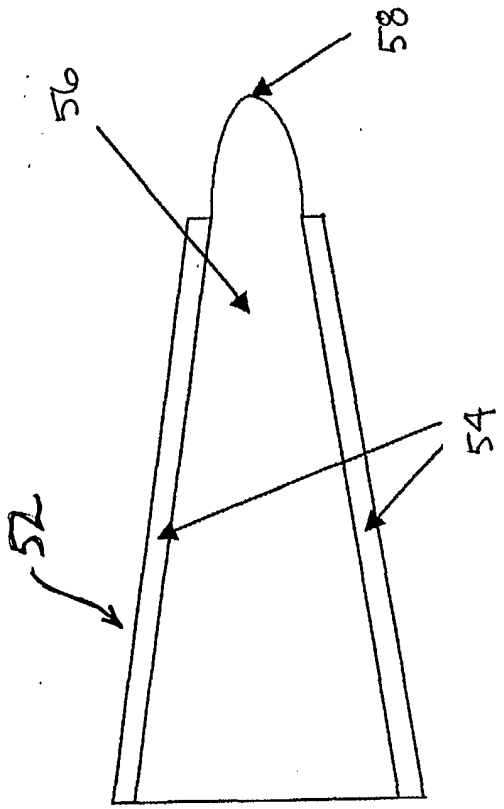


FIG. 13

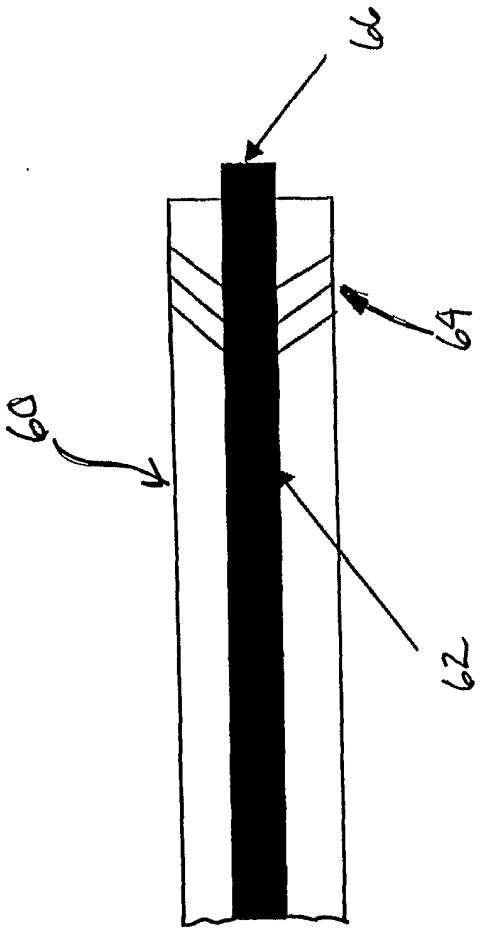


FIG. 14

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US01/05359

A. CLASSIFICATION OF SUBJECT MATTER

IPC(7) :H01J 35/00
US CL :378/122

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 378/122

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

NONE

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

NONE

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 5,854,822 A (CHORNENKY et al) 29 December 1998 (29.12.1998), see the entire document.	1-24
A,P	US 6,148,061 A (SHEFER et al) 14 November 2000 (14.11.2000), see the entire document.	1-45

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

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

Date of the actual completion of the international search 02 MAY 2001	Date of mailing of the international search report 12 JUN 2001
Name and mailing address of the ISA/US Commissioner of Patents and Trademarks Box PCT Washington, D.C. 20231 Facsimile No. (703) 305-3230	Authorized officer Craig Church Telephone No. (703) 308-4861

Form PCT/ISA/210 (second sheet) (July 1998)★