



(11) **EP 2 111 165 B1**

(12) **EUROPEAN PATENT SPECIFICATION**

(45) Date of publication and mention of the grant of the patent:  
**07.06.2017 Bulletin 2017/23**

(51) Int Cl.:  
**A61B 8/12 (2006.01) A61B 1/04 (2006.01)**  
**A61B 1/07 (2006.01) G01D 5/347 (2006.01)**

(21) Application number: **08706237.8**

(86) International application number:  
**PCT/CA2008/000089**

(22) Date of filing: **21.01.2008**

(87) International publication number:  
**WO 2008/086613 (24.07.2008 Gazette 2008/30)**

(54) **IMAGING PROBE WITH COMBINED ULTRASOUND AND OPTICAL MEANS OF IMAGING**

**BILDGEBUNGSSONDE MIT KOMBINIERTEM ULTRASCHALL UND OPTISCHEM BILDGEBUNGSMITTEL**

**SONDE D'IMAGERIE DOTÉ D'UN MOYEN ULTRASONIQUE ET OPTIQUE D'IMAGERIE**

(84) Designated Contracting States:  
**AT BE BG CH CY CZ DE DK EE ES FI FR GB GR HR HU IE IS IT LI LT LU LV MC MT NL NO PL PT RO SE SI SK TR**

- **YANG, Victor, Xiao Dong**  
**Toronto, Ontario M5B 2C2 (CA)**
- **FOSTER, Francis, Stuart**  
**Toronto, Ontario M4E 1Y4 (CA)**

(30) Priority: **19.01.2007 US 881169 P**

(74) Representative: **Denemeyer & Associates S.A.**  
**55, rue des Bruyères**  
**1274 Howald (LU)**

(43) Date of publication of application:  
**28.10.2009 Bulletin 2009/44**

(73) Proprietor: **SUNNYBROOK HEALTH SCIENCE CENTRE**  
**Toronto, Ontario M4N 3M5 (CA)**

(56) References cited:  
**EP-A2- 1 972 262 WO-A1-00/33120**  
**WO-A1-2007/123518 WO-A2-2008/057573**  
**DE-A1- 2 950 203 JP-A- 2004 290 548**  
**US-A- 5 469 853 US-A- 5 916 210**  
**US-A1- 2005 075 574 US-A1- 2005 101 859**  
**US-A1- 2006 058 614 US-A1- 2006 116 571**  
**US-A1- 2006 287 595 US-A1- 2007 282 197**  
**US-A1- 2009 088 648 US-B1- 6 264 610**  
**US-B1- 6 390 978 US-B1- 6 690 958**

(72) Inventors:

- **COURTNEY, Brian**  
**Toronto, Ontario M4V 2E9 (CA)**
- **MUNCE, Nigel, Robert**  
**Hamilton, Ontario L8P1T2 (CA)**
- **THIND, Amandeep Singh**  
**Toronto, Ontario M4S 2J7 (CA)**

**EP 2 111 165 B1**

Note: Within nine months of the publication of the mention of the grant of the European patent in the European Patent Bulletin, any person may give notice to the European Patent Office of opposition to that patent, in accordance with the Implementing Regulations. Notice of opposition shall not be deemed to have been filed until the opposition fee has been paid. (Art. 99(1) European Patent Convention).

**Description****FIELD OF THE INVENTION**

[0001] The present invention relates generally to the field of imaging mammalian tissues and structures using high resolution imaging using combined high frequency ultrasound (IVUS) and optical imaging methods such as optical coherence tomography (OCT) and to accurate co-registering of images obtained from ultrasound image signals and optical image signals during scanning a region of interest.

**BACKGROUND OF THE INVENTION**

[0002] High resolution imaging of the interior of the body (or for dermatologic or ophthalmology applications not restricted to the interior) serves multiple purposes, including any of i) assessing tissue structures, anatomy and composition ; ii) planning and / or guiding interventions on localized regions of the body; and iii) assessing the result of interventions that alter the structure, composition or other properties of the localized region. High resolution imaging in this particular case refers to high frequency ultrasound and optical imaging methods. For the purposes of this invention, high frequency ultrasound typically refers to imaging with frequencies of greater than 3 MHz, and more typically in the range of 9 to 100 MHz.

[0003] High frequency ultrasound is very useful for intravascular and intracardiac procedures. For these applications, the ultrasound transducers are incorporated into a catheter or other device that can be inserted into the body. By way of example, two particularly important implementations of high frequency ultrasound are intravascular ultrasound (IVUS), for imaging blood vessels, and intracardiac echocardiography (ICE) for imaging cardiac chambers. Both ICE and IVUS are minimally invasive, and involve placing one or more ultrasound transducers inside a blood vessel or cardiac chamber to take high quality images of these structures.

[0004] Optical imaging methods based on fiber optic technology used in the field of medicine include optical coherence tomography (OCT), angiography, near infrared spectroscopy, Raman spectroscopy and fluorescence spectroscopy. These modalities typically require the use of one or more optical fibers to transmit light energy along a shaft between an imaging site and an imaging detector. Optical coherence tomography is an optical analog of ultrasound, and provides imaging resolutions on the order of 1 to 30 microns, but does not penetrate as deeply into tissue as ultrasound in most cases. Fiber optics can also be used to deliver energy for therapeutic maneuvers such as laser ablation of tissue and photodynamic therapy.

[0005] Additional forms of imaging related to this invention include angiography, endoscopy and other similar imaging mechanisms that involves imaging a site inside

the patient using a probe to take pictures based on the back-reflection of light.

[0006] High resolution imaging means have been implemented in many forms for assessing several different regions of mammalian anatomy, including the gastrointestinal system, the cardiovascular system (including coronary, peripheral and neurological vasculature), skin, eyes (including the retina), the genitourinary systems, breast tissue, liver tissue and many others. By way of example, imaging of the cardiovascular system with high frequency ultrasound or optical coherence tomography has been developed for assessing the structure and composition of arterial plaque.

[0007] High-resolution imaging has been used to measure vessel or plaque geometry, blood flow through diseased arteries, the effects of interventions on arterial plaque (such as by atherectomy, angioplasty and/or stenting). Attempts have also been made using high resolution imaging to identify vascular lesions that have not led to clinical symptoms, but are at increased risk of rupturing or eroding and causing an acute myocardial infarction. These so-called "vulnerable plaques" are an area of interest as the prospect of treating such plaques to pre-empt adverse clinical events is conceptually appealing.

[0008] Chronic total occlusions are a specific subset of vascular lesions where the entire lumen of the vessel has been occluded (based on the angiographic appearance of the lesion) for over approximately one month. Most intravascular imaging modalities are "side-viewing" and require passage of an intravascular imaging device through a lesion. In order to image chronic total occlusions, methods of high resolution imaging would be more useful if they were adapted to a "forward-looking" rather than "side-viewing" configuration.

[0009] Several of these high resolution imaging means are dependent on the use of a rotary shaft to transmit torque to an imaging device near the distal end of the probe. These rotary shafts are often long, thin and flexible, such that they can be delivered through anatomical conduits, such as the vasculature, genitourinary tracts, respiratory tracts and other such bodily lumens. Ideally, when a continuous torque is applied to the cable in a specified direction the torque cable develops a property of having a close relation between the degree of rotation at its proximal and distal ends. This allows the simplification of the design of the ultrasound catheter by making the angle of rotation at the distal end of the torque cable (within the body) a reasonable approximation of the angle of rotation at the proximal end of the torque cable (outside of the body).

[0010] The rotation of the torque cable or shaft at the point from which the imaging occurs may not be identical to the rotation occurs at the proximal end of the torque cable or shaft. This occurs especially when the flexible shaft is delivered through tortuous passageways and is, at least in part, due to inertia and friction between the rotating components and stationary components of the

imaging shaft. The assumption that the rotational speed of the proximal and distal ends of the rotary shaft are equal to each other is also less likely to be valid if the rotational speed varies over time. The undesirable result of not knowing the true angular velocity of the imaging probe at the point from which the imaging beam is directed towards the tissue leads to an artifact referred to as non-uniform rotational distortion (NURD). NURD can lead to significant distortion of the image and a concomitant reduction in the geometric accuracy of the image. Knowledge of a more precise estimation of the true rotary speed of the distal rotary shaft or an imaging assembly attached to the rotary shaft can help overcome such distortion by providing more accurate information for image reconstruction. A better estimation of the rotary speed can also help improve the accuracy of co-registration of images when more than one imaging modality is implemented on the imaging probe (such as combined ultrasound and optical imaging).

**[0011]** While use of more than one type of imaging technique, such as ultrasound and optical techniques, have both proved valuable in medical applications for high resolution imaging, they are not commonly used in tandem. As described in the Summary of the related art below, there are some designs that exist for the combination of optical and ultrasound technologies. However, the limitations in these designs have prevented their acceptance.

**[0012]** Namely, designs that incorporate optical and ultrasound technologies offset the ultrasound and optical imaging mechanisms, such as disclosed in (Maschke, US patent 7289842 resulting in the acquisition of unaligned ultrasound and optical signals. Alignment of the resultant data from these two imaging means requires movement of the imaging mechanisms and is prone to registration errors due to (i) non-uniform rotational distortion (NURD), (ii) motion of the object occurring between successive imaging of the same location using the two imaging means, (iii) variability in the object being imaged, and (iv) difficulty in accurately tracking the location of the imaging means. All these effects result in inaccurate co-registration which limits the usefulness of the acquisition of data from the two imaging means.

### Summary of the related art

**[0013]** A catheter-based system for intravascular ultrasound is described by Yock (U.S. Patent No. 4,794,931) to provide high resolution imaging of structures in blood vessels. This system comprises an outer sheath, within which there is an ultrasound transducer near the distal end of a long torque cable. When a motor rotates the torque cable and ultrasound transducer assembly, 2D cross-sectional images of anatomical structures, such as blood vessels, can be made. Linear translation of the catheter or the torque cable and ultrasound transducer in combination with the rotational motion of the ultrasound transducer allows for acquisition of a series of 2D

images along the length of the catheter.

**[0014]** Milo et al (U.S. Patent No. 5,429,136) and Lenker et al (U.S. Patent Nos. 6,110,121 and 6,592,526) describe reciprocating and vibrating means for scanning an ultrasound imaging beam in circumferential or longitudinal directions at the end of the catheter. Reciprocating or vibrating means obviates the need to use a mechanism such as a slip ring to provide an electrical connection to a probe that rotates more than a few rotations in a particular direction, such as more than one or two rotations. Similarly, certain implementations of optical imaging can avoid the use of optical rotary joints using reciprocating or vibrating means.

**[0015]** Liang et al. (United States Patent Nos. 5,606,975 and 5,651,366) describe means of implementing forward-looking intravascular ultrasound where ultrasound is directed towards a mirror that causes the ultrasound beam to propagate at an angle from the longitudinal axis of a rotating torque cable advanced within the vasculature. Liang et al. also describe means of varying the angle of deflection of the mirror using either a micro-motor, a gear clutch mechanism, steering cables or bimorph elements such as shape memory alloys, piezoelectric files or conductive polymers. Figure 13 of U.S. Patent No. 5,651,366 shows a diagram of a forward looking ultrasound probe combined with a fiber optic to deliver laser ablation energy via a fiber and mirror in a coaxial direction to the ultrasound imaging beam, but does not relate to combined optical and acoustic imaging or provide for optical focusing elements which would be of benefit for imaging purposes.

**[0016]** The use of intravascular ultrasound (IVUS) has since become commonplace, with many improvements and adaptations to the technology. A flexible torque cable (Crowley, U.S. Patent No. 4,951,677) improves the fidelity of the transmission of rotational torque along the length of an IVUS catheter, minimizing an artifact known as non-uniform rotational distortion.

**[0017]** The center frequency of IVUS lies within the range of 3 to 100 MHz and more typically in the range of 20 to 50 MHz. Higher frequencies provide higher resolution but result in worse signal penetration and thus a smaller field of view. Depth of penetration can range from less than a millimeter to several centimeters depending on several parameters such as center frequency and geometry of the transducer, the attenuation of the media through which the imaging occurs and implementation-specific specifications that affect the signal to noise ratio of the system.

**[0018]** Variations of high frequency ultrasound exist, where the signal acquisition and / or analysis of the back-scattered signal is modified to facilitate obtaining or inferring further information about the imaged tissue exist. These include elastography, where the strain within tissue is assessed as the tissue is compressed at different blood pressures (de Korte et al Circulation. 2002 Apr 9;105(14):1627-30); Doppler imaging which assesses motion such as blood flow within anatomic structures;

virtual histology, which attempts to infer the composition of tissue using the radio-frequency properties of the back-scattered signal combined with a pattern recognition algorithm (Nair, U.S. Patent No.6,200,268); second harmonic imaging (Goertz et al, Invest Radiol. 2006 Aug;41(8):631-8) and others. Each of these forms of imaging can be improved upon by means described in the present invention.

**[0019]** Ultrasound transducers themselves are improving considerably, including the use of single crystal ultrasound transducers and composite ultrasound transducers.

**[0020]** Hossack et al (WO/2006/121851) describe a forward looking ultrasound transducer using a CMUT transducer and a reflective surface.

**[0021]** Tearney et al (U.S. Patent No. 6,134,003) describe several embodiments that enable optical coherence tomography to provide higher resolution imaging than is readily obtained by high frequency ultrasound or IVUS.

**[0022]** Boppart et al (United States Patent No. 6,485,413) describe several embodiments of optical coherence tomography imaging, including forward-looking implementations. Either an optical fiber or a gradient index (GRIN) lens are displaced using a mechanism such as a motor, a piezoelectric, a moveable wire, inflation means and others.

**[0023]** Mao et al (Appl Opt. 2007 Aug 10;46(23):5887-94) describe methods for creating ultrasmall OCT probes using single mode fiber, coupled to a small length of GRIN fiber which acts as a lens. Including an optical spacer between the fiber and the lens can alter the working distance of the fiber-lens system. Furthermore, adding a small length of no-clad fiber to the distal end, and cutting the no-clad fiber at an angle can add a deflecting element to the end of the fiber-lens system. This deflecting element enables side-viewing imaging, which could also be accomplished using a small prism or mirror.

**[0024]** Variations of optical coherence tomography (OCT) include polarization sensitive OCT (PS-OCT) where the birefringent properties of tissue components can be exploited to obtain additional information about structure and composition; spectroscopic OCT which similarly provides improved information regarding the composition of the imaged structures; Doppler OCT which provides information regarding flow and motion; elastography via OCT; and optical frequency domain imaging (OFDI), which allows for a markedly more rapid acquisition of imaging data and therefore enables imaging to occur over a larger volume of interest in less time. Again, each of these forms of imaging can be improved upon by means of the present invention.

**[0025]** Several other forms of fiber-optic based imaging exist other than OCT. Amundson et al describe a system for imaging through blood using infra-red light (United States Patent No. 6,178,346). The range of the electromagnetic spectrum that is used for their imaging

system is selected to be one which optimizes penetration through blood, allowing optical imaging through blood similar to that afforded by angioscopy in the visible spectrum, but without the need to flush blood away from the region being imaged.

**[0026]** Dewhurst (US Patent No. 5,718,231) discloses a forward looking probe for intravascular imaging where a fiber optic travels through an ultrasound transducer to shine light on a target tissue straight in front of the end of the probe. The light then interacts with the target tissue and makes ultrasound waves, which are received by the ultrasound sensor and the images are photoacoustic images only as the system is not configured to receive and process optical images. The ultrasound sensor used in the Dewhurst device is limited to thin film polymeric piezoelectrics, such as thin film PVDF, and is used only to receive ultrasound energy, not to convert electrical energy to ultrasound.

**[0027]** Angioscopy, endoscopy, bronchoscopy and many other imaging devices have been described which allow for the visualization of internal conduits and structures (such as vessels, gastrointestinal lumens and the pulmonary system) in mammalian bodies based on the principle of illuminating a region within the body near the distal end of a rigid or flexible shaft. Images are then created by either having a photodetector array (such as a CCD array) near the end of the shaft or by having a bundle of fiber optics transmit the received light from the distal end of the shaft to the proximal end where a photodetector array or other system that allows the operator to generate or look at an image representative of the illuminated region. Fiber bundles are bulky and reduce the flexibility of the shaft among other disadvantages.

**[0028]** Other fiber optic based modalities for minimally invasive assessment of anatomic structures include Raman spectroscopy as described by Motz et al (J Biomed Opt. 2006 Mar-Apr; 11 (2)), near infrared spectroscopy as described by Caplan et al (J Am Coll Cardiol. 2006 Apr 18;47(8 Suppl):C92-6) and fluorescence imaging, such as tagged fluorescent imaging of proteolytic enzymes in tumors (Radiology. 2004 Jun;231(3):659-66).

**[0029]** The ability to combine ultrasound and optical coherence tomography onto a single catheter would be extremely advantageous. Kubo et al presented an interesting in vivo study of coronary arteries using OCT, IVUS and angioscopy to assess the morphology of lesions that have caused an acute myocardial infarction (Journal of American College of Cardiology, Sept 4, 2007, 10(50):933-39). They demonstrate that there are benefits to imaging with each of these modalities. However, in order to execute their study, they had to use separate catheters for each of IVUS, OCT and angioscopy imaging modalities as no catheters that combine these functions have been commercialized to date. Kawasaki et al previously compared OCT, conventional IVUS and a variant of IVUS known as integrated backscatter IVUS on cadaveric specimens of coronary arteries using separate probes for the OCT and IVUS components. Brezinski et

al (Heart. 1997 May; 77(5):397-403) had previously demonstrated ex vivo studies on dissected aortic specimens where IVUS and OCT images were compared, again using separate probes. The OCT probes in this latter study were not suitable for in vivo use.

**[0030]** Optical coherence tomography generally has superior resolution to ultrasound and has the potential to better identify some structures or components in vascular and other tissues than ultrasound. For example, fibrous cap thickness or the presence of inflammatory or necrotic regions near the surface of arteries may be better resolved with optical coherence tomography. However, optical coherence tomography is limited by its small penetration depth (on the order of 500 to 3000 microns) in most biologic media. Most such media are not optically transparent.

**[0031]** Meanwhile, ultrasound has the ability to better penetrate through biological media such as blood and soft tissues and has a depth of penetration that typically extends several millimeters or centimeters beyond that of optical coherence tomography. The ability to image with either or both methods of imaging using a combined imaging device provides advantages with respect to selecting the required resolution and depth of penetration. Furthermore, much of the information acquired by optical coherence tomography is complementary to that acquired by ultrasound and analysis or display of information acquired by both imaging methods would improve the ability to better understand the interrogated tissue, such as with respect to its composition.

**[0032]** These differences between IVUS and OCT are well known in the art. Maschke (United States Patent Publication No. 2006/0116571 corresponding to U.S. Patent Application Serial No. 11/291,593) describes an embodiment of a guidewire with both OCT and IVUS imaging transducers mounted upon it. The described invention has several shortcomings. Guidewires are typically 0.014" to 0.035" in diameter (approximately 350 microns to 875 microns), yet ultrasound transducers typically are at least 400 microns x 400 microns and generally are larger in size for the frequencies in the 20 to 100 MHz range. If the transducer is too small, the beam is poorly focused and has poor signal properties. In Maschke the IVUS and OCT imaging mechanisms are located at different positions along the length of the guidewire and a drawback to this type of configuration having the IVUS and OCT imaging means located at different positions along the length of an imaging shaft does not allow for optimal co-registration of images.

**[0033]** U.S. Patent No. 7,289,842 issued to Maschke describes an imaging system that combines IVUS and OCT on a catheter where the IVUS and OCT imaging elements are longitudinally displaced from each other along the length of a catheter that rotates around its longitudinal axis. Maschke also describes generating images where the center portion of the images are substantially derived from the output of the higher resolution OCT imaging portion of the system while the outer portion of

the images are substantially derived from the output of the ultrasound imaging portion of the system, to make use of ultrasound's greater depth of penetration in combination with OCT's higher resolution for tissues close to the catheter.

**[0034]** Park et al (U.S. Patent application 11/415,848) also briefly refers to the notion of having a catheter that combines IVUS and OCT imaging onto a single catheter.

**[0035]** However, the integration of means for combined acoustic and optical imaging, such as combined IVUS and OCT imaging, onto a single device is not trivial. Having an optical imaging element and an acoustic imaging element longitudinally separated from each other on a primarily rotating catheter does not provide an ideal configuration for combined imaging. A more ideal configuration would enable the acquisition of high quality acoustic and optical signals from which ultrasound and optical-based images could be made while enabling the acoustic and optical images to be registered with each other in a highly precise manner.

**[0036]** For example, by simply placing an IVUS imaging element in line with an OCT imaging element along the length of the catheter, the center of the imaging planes of the IVUS and OCT images will be separated from one another by a distance of at least approximately half the length of the ultrasound transducer and half the length of the optical imaging elements.

**[0037]** Mechanical IVUS transducers for vascular imaging are typically more than 400 microns in length. The separation between the IVUS and OCT planes of imaging in a configuration such as that proposed by Maschke would require at least 250 microns of separation between the optical and acoustic imaging planes. Typically, mechanical IVUS rotates at 30 frames per second with a pullback rate of 0.5 mm/s, meaning that from a given time point to, at least 15 imaging frames or 500 milliseconds would elapse between the time that the more distally placed imaging means would translate to the same position at which the more proximally placed imaging means was originally positioned at time  $t_0$ . This separation of several hundred milliseconds or several rotations of the imaging probe makes it difficult to precisely register the imaging data from one imaging means with the other.

**[0038]** This is particularly relevant given the fact that the catheter can undergo significant unintentional lateral and longitudinal displacements within body lumen in that time period, such as those displacements that occur as a result of cardiac contraction and pulsatile flow. Non-uniform rotational distortion (NURD) can also have an impact on the ability to accurately register images acquired several rotations apart from each other. Any imprecision of the registration of the two data sets is even more significant when one considers the scale at which important pathologies, such as vulnerable plaques can be found. Dramatic differences in the appearance of an arterial plaque's composition (e.g. the thickness of a fibrous cap, the presence of a calcified nodule or the extent of an atheromatous deposit) can be observed in as little

as a few hundreds microns along the length of a vessel. Similarly, small but potentially relevant sidebranches of anatomic conduits, such as blood vessels, can have dimensions on the order of less than a hundred microns.

**[0039]** Previous experiments and implementations of IVUS and OCT or other combinations of acoustic and optical imaging have not been provided that enable significant precision in the registration of the imaging data from the two or more imaging means in a manner that is suitable for minimally invasive imaging, such as intravascular imaging.

**[0040]** To the best of our knowledge, previous experiments and implementations of IVUS and OCT or other combinations of acoustic and optical imaging have not been provided that enable significant precision in the registration of the imaging data from the two or more imaging means in a manner that is suitable for minimally invasive imaging, such as intravascular imaging.

**[0041]** It would be very advantageous to also provide high resolution imaging probes that combine acoustic and optical imaging onto "forward-looking" probes rather than "side-viewing" probes. It would also be helpful to provide similar probes that can look backwards, or from multiple angles in a generally side-viewing configuration.

**[0042]** It would also be advantageous to provide high-resolution imaging probes that combine ultrasound imaging with one or more optical imaging means.

**[0043]** It would also be advantageous to provide minimally invasive imaging probes that can be used for photoacoustic imaging or sonoluminescent imaging.

**[0044]** It would also be advantageous to provide minimally invasive imaging means where one of the imaging means provides helpful information regarding the direction in which the other imaging means is acquiring imaging data.

**[0045]** Document US6390978 B1 may be considered to disclose an imaging probe for insertion into bodily lumens and cavities for imaging an interior of said bodily lumens and cavities, comprising:

a) an elongate shaft having a longitudinal axis having distal and proximal end sections and an elongate midsection, an imaging assembly being located on said elongate hollow shaft remote from said proximal end section, said imaging assembly being connected to a first end of an imaging conduit, said imaging conduit extending through the elongate hollow shaft and being connected at a second end thereof to an image processing and display system through the proximal end section, said imaging conduit including a fiber optic having a distal end and said imaging assembly including an optical emitter / collector including light directing and receiving means associated with said distal end of a fiber optic for directing light imaging energy out of a distal end of said fiber optic and receiving reflected light imaging energy signals and directing said received reflected light imaging energy signals back to said image processing

and display system, said imaging assembly including an ultrasound transducer and said ultrasound transducer being adapted to emit and receive ultrasound imaging energy and said imaging conduit including an electrical conductor for electrically coupling the ultrasound transducer to an ultrasound signal generator and said image processing and display system; wherein said light directing and receiving means and at least part of said distal end of said fiber are in, or are within, or are located in/within, or are contained in/within, or reside in/within said transducer; and are encompassed, or surrounded, or enclosed by said transducer to an extent permitting, or allowing, or providing for the distal end of said fiber to enter said transducer; and are not located in said ultrasound transducer substrate and wherein both the acoustic and optical signals are emitted at the same angle relative to the longitudinal axis of the probe and wherein the optical signals are emitted through a single optically transmissive channel in said transducer; and

d) a controller being connected to the image processing and display system, and being configured to process the images obtained from ultrasound imaging and optical imaging during scanning of a region of interest.

#### SUMMARY OF THE INVENTION

**[0046]** The present invention provides embodiments of imaging probes for combining acoustic and optical imaging means in a manner that facilitates simultaneous imaging by two or more imaging methods. The embodiments enable methods to accurately co-register the images obtained from each of the modalities. In some embodiments, the current invention provides embodiments for combining acoustic imaging means with the delivery of therapeutic energy, such as ultraviolet light for photodynamic therapy or laser energy for ablation procedures.

**[0047]** The present invention also provides embodiments where one form of imaging is used to help with the reconstruction of the second form of imaging. This is more specifically related to monitoring the position or orientation of a component in the image probe that subsequently determines the position or orientation of the imaged region.

**[0048]** The present invention provides methods for combining high frequency ultrasound and optical coherence tomography into a combined imaging system.

**[0049]** The present invention provides novel means for implementing a combined ultrasound and optical imaging system where the volume scanned includes a region either forward of, or behind, the location of the imaging transducers.

**[0050]** The present invention provides the ability to take images similar to those produced by angioscopy, endoscopy and similar imaging techniques using a single optic or a small number of fiber optics, in combination

with means to acquire ultrasound images. These optical images can also be acquired using infrared and / or visible wavelengths.

**[0051]** The present invention provides means for combining high frequency ultrasound and optical coherence tomography where the volumes scanned include regions either forward of, or behind, the locations of the imaging transducers.

**[0052]** Embodiments of the present invention are able to scan a region for the purposes of imaging or delivery of therapeutic energy the region accessed by a shaft where changes in the rotation velocity of the shaft causes changes in the direction of either an emitter and / or receiver of acoustic and / or optical energy.

**[0053]** The present invention also facilitates certain forms of high resolution imaging that use acoustic energy to create optical energy (sonoluminescence imaging) or optical energy to create acoustic energy (photoacoustic imaging). The invention is defined in claim 1, with further aspects in the dependent claims. A further understanding of the functional and advantageous aspects of the invention can be realized by reference to the following detailed description and drawings.

#### BRIEF DESCRIPTION OF THE DRAWINGS

**[0054]** Preferred embodiments of the invention will now be described, by way of example only, with reference to the drawings, in which:

**Figure 1** is a schematic of an imaging system including ultrasound and optical imaging components;

**Figure 2** is a perspective drawing of a flexible imaging probe with an adapter, conduit and imaging assembly;

**Figure 2a** is a cross sectional view of the mid section of the imaging probe of Figure 2 taken along the dotted line;

**Figure 2b** is an expanded perspective drawing of the distal region of the imaging probe of **Figure 2**;

**Figure 2c** shows a schematic of how the rotary and non-rotary components of the imaging probe can be coupled with an adapter to the rest of an imaging system.

**Figure 2d** is a perspective drawing of an example of the coupling of the rotary and non-rotary components of the probe to an adapter.

**Figures 3a to 3e** are representative of general imaging catheter configurations described in the prior art;

**Figure 3a** shows one embodiment of an over-the-wire configuration for an external sheath that may be incorporated with the imaging probe if a guidewire lumen is included;

**Figure 3b** shows a cross-section through the imaging probe to demonstrate the guidewire lumen configuration.

**Figure 3c** shows a rapid access configuration for an

external sheath that may be incorporated with the imaging probe if a guidewire lumen is included;

**Figure 3d** shows a cross-section through a portion of the imaging probe that does not contain a guidewire lumen;

**Figure 3e** shows a cross-section through a portion of the imaging probe that does contain a guidewire lumen;

**Figures 4a to 4l** are examples of ultrasound transducers that contain a hole for allowing transmission of optical energy through the transducer that enables optical and acoustic imaging of regions that are precisely aligned with each other, as well as means to deflect the path of the imaging light;

**Figures 5a to 5f** are examples of ultrasound transducers that contain a hole for allowing transmission of optical energy through the transducer that enables optical and acoustic imaging of regions that are precisely aligned with each other, without a means to deflect the path of the imaging light;

**Figures 6a to 6c** demonstrate representative acoustic transducer configurations, with **Figure 6a** not having a hole in the transducer. **Figures 6d to 6f** demonstrate representative simulation results of the effects of placing a hole through an ultrasound transducer on the acoustic beam pattern produced by the ultrasound transducer with **Figure 6d** not having a hole;

**Figures 7a to 7e** show examples of ultrasound transducers that have an optical apparatus for transmitting and / or receiving optical imaging energy either on top of or recessed within an acoustic transducer;

**Figure 8a** is a perspective view of an imaging assembly suitable for side viewing with both acoustic and optical imaging;

**Figures 8b** is a side view of the imaging assembly in **Figure 8a**;

**Figures 8c to 8e** are end views of the imaging assembly in **Figure 8a** in different rotated positions;

**Figures 9a to 9c** depict configurations whereby an optical imaging emitter / receiver is embedded into the backing material **435** of an acoustic transducer.

**Figures 10a to 10e** are similar to **Figures 8b to 8e** showing the imaging assembly being rotated in a reciprocating fashion rather than in a single rotational direction;

**Figure 11** shows a perspective view of an imaging probe where the predominant motion is a longitudinal motion where the surface swept by the optical beam and the acoustic beam are two co-planar rectangles;

**Figure 12** shows a perspective view of an embodiment of an imaging probe where the optical imaging system is configured such that the optical imaging beams are angled such that these imaging beams substantially converge or overlap;

**Figure 13** is a cross sectional view of an imaging assembly suitable for side viewing with both acoustic and optical imaging;

**Figures 14a** is a cross sectional view of an imaging assembly suitable for forward viewing with both acoustic and optical imaging;

**Figures 14b** is a cross sectional view of an imaging assembly suitable for forward viewing with both acoustic and optical imaging in which an artificial muscle polymer can be used to deform the distal region of the imaging probe;

**Figures 15a** is a cross sectional view of an imaging assembly suitable for side viewing with both acoustic and optical imaging using a reflective component to direct the optical and acoustic beams in the sideways direction;

**Figures 15b** and **15c** are similar to **Figure 15a** but in which the reflective component is mounted about a pivot point so the optical and acoustic beams can be scanned in the sideways direction at a variable angle;

**Figure 16a** is a cross section of an embodiment of an imaging probe using a tiltable component where the tilting action is modulated by centripetal acceleration due to the rotational motion of the imaging assembly around the longitudinal axis;

**Figure 16b** is a view along the line **16b-16b** of **Figure 16a**;

**Figure 16c** is a cross section of the imaging probe of **Figure 16a** but with the tiltable component at a different angle during use;

**Figure 16d** is a view along the line **16d-16d** of **Figure 16c**;

**Figure 17a** is a perspective drawing of a deflecting component that comprises a flat optically reflective layer and a shaped acoustically reflective layer;

**Figures 17b** through **17d** depict cross-sections of the deflecting component of **Figure 17a**;

**Figure 18a** is a perspective view of an ultrasound imaging transducer with two (2) optical imaging emitters / receivers through two (2) separate optically transmissive channels in the acoustic transducer;

**Figures 18b** is a perspective view of an embodiment of an imaging probe having an ultrasound imaging transducer with two (2) optical imaging emitters / receivers arranged in a manner such that they are aligned with the predominant rotary motion of the imaging assembly;

**Figure 18c** is a view along arrow C of **Figure 18b**;

**Figure 19** is a schematic of a system where there are two optical imaging systems that are coupled to the same optical imaging waveguide via optical routing circuitry;

**Figures 20a** and **20b** demonstrate sector-shaped patterns for simultaneously demonstrating portions of two (2) or more images that are co-registered with each other;

**Figures 21 a** and **21 b** demonstrate arbitrary patterns for simultaneously showing portions of 2 or more images that are co-registered with each other;

**Figure 22** is a schematic of a display which transi-

tions over time from one image to another, co-registered image;

**Figures 23a** and **23b** demonstrate how a feature in a first image can be mapped onto a feature in another image that is co-registered with the first image;

**Figures 24a** and **24b** demonstrate how a contour feature in a first image can be mapped into another image with is co-registered with the first image and vice versa; and

**Figures 25a** and **25b** provide a schematic for how a composite image can be constructed from two (2) or more co-registered imaging datasets.

## DETAILED DESCRIPTION OF THE INVENTION

**[0055]** Without limitation, the majority of the systems described herein are directed to an imaging probe that enables imaging by both optical and acoustic means. As required, embodiments of the present invention are disclosed herein. However, the disclosed embodiments are merely exemplary. The Figures are not to scale and some features may be exaggerated or minimized to show details of particular elements while related elements may have been eliminated to prevent obscuring novel aspects. Therefore, specific structural and functional details disclosed herein are not to be interpreted as limiting but merely as a basis for the claims and as a representative basis for teaching one skilled in the art to variously employ the present invention. The illustrated embodiments are directed to an imaging probe that enables imaging by both optical and acoustic means.

**[0056]** As used herein, the term "about", when used in conjunction with ranges of dimensions, temperatures or other physical properties or characteristics is meant to cover slight variations that may exist in the upper and lower limits of the ranges of dimensions so as to not exclude embodiments where on average most of the dimensions are satisfied but where statistically dimensions may exist outside this region. For example, in embodiments of the present invention dimensions of components of an imaging probe are given but it will be understood that these are not meant to be limiting.

**[0057]** As used herein, the phrase "co-registration of images" refers to the process of identifying a subset of imaging data acquired by one imaging means with a subset of imaging data acquired using another imaging means where the identified imaging data from the two means was acquired by detecting a form of imaging energy (e.g. photons or ultrasound) from the same object (or tissue in the case of the present invention). Each co-registered point in the first subset can then be mapped to a corresponding point in the second subset such that the two points from the two different imaging means are thought to have been acquired from a similar focal region of the imaged object (or tissue).

**[0058]** Successful and accurate co-registration of images, or portions thereof, between images acquired using two (2) or more imaging means is helpful in that it can



provide multiple opportunities to assess features of interest of the imaged object by more than one imaging means.

**[0059]** **Figure 1** represents an overview of an exemplary imaging system constructed in accordance with the present invention shown generally at **10**. It comprises an imaging probe **12**, which connects via an adapter **14** to an image processing and display system **16**. The image processing and display system **16** comprises the necessary hardware to support one or more of the following imaging modalities: 1) ultrasound, 2) optical coherence tomography, 3) angioscopy, 4) infrared imaging, 5) near infrared imaging, 6) Raman spectroscopy-based imaging and 7) fluorescence imaging.

**[0060]** Implementations of the optical coherence tomography, ultrasound, angioscopy and infrared imaging circuitry have been described in the prior art.

**[0061]** The system herein described further typically comprises a controller and processing unit **18** to facilitate the coordinated activity of the many functional units of the system, and may further comprise a display and/or user interface and may further comprise electrode sensors to acquire electrocardiogram signals from the body of the patient being imaged. The electrocardiogram signals may be used to time the acquisition of imaging data in situations where cardiac motion may have an impact on image quality. The optical circuits and electronics **21** forming image processing and display system, if included in a particular implementation of the present disclosure, may include any or all of the following components: interferometer components, one or more optical reference arms, optical multiplexors, optical demultiplexors, light sources, photodetectors, spectrometers, polarization filters, polarization controllers, timing circuitry, analog to digital converters and other components known to facilitate any of the optical imaging techniques described in the background and prior art sections. The ultrasound circuitry **20** may include any or all of the following components: pulse generators, electronic filters, analog to digital converters, parallel processing arrays, envelope detection, amplifiers including time gain compensation amplifiers and other components known to facilitate any of the acoustic imaging techniques described in the background and prior art sections.

**[0062]** The controller and processing units **18**, if included in a particular implementation of the present disclosure, serve multiple purposes and the components would be markedly adapted based on the needs of a particular imaging system. It could include one or a combination of motor drive controller, data storage components (such as memory, hard drives, removable storage devices, readers and recorders for portable storage media such as CDs and DVDs), position sensing circuitry, timing circuitry, cardiac gating functionality, volumetric imaging processors, scan converters and others. A display and user interface **22** is also optionally provided for either real time display or display of data at a time later than the time at which imaging data is acquired.

**[0063]** The imaging probe **12** comprises an imaging assembly **30** near its distal end **32**, an optional conduit **34** along a substantial portion of its length, and a connector **36** at its proximal end **38**. For the purposes of this disclosure, an imaging assembly **30** generally refers to the component of the imaging probe **12** from which the signals (acoustic or optical (or both)) are collected for the purposes of imaging a region that is proximate to the imaging assembly **30**. The imaging assembly **30** includes at least one or more emitters of imaging energy and at least one or more receivers of imaging energy. For the purposes of this disclosure, "imaging energy" refers to both light and acoustic energy. Specifically, light refers to electromagnetic waves that span the ultraviolet, visible and infrared spectrum of wavelengths. For example, for acoustic imaging, the imaging assembly **30** contains an ultrasound transducer that is both an emitter and receiver of acoustic energy.

**[0064]** For optical imaging, the imaging assembly **30** typically contains the distal tip of a fiber optic, as well as a combination of optical components such as a lens (such as a ball lens or GRIN lens), which collectively serve the purpose of acting as an optical receiver and may also serve as an optical emitter. A mirror and/or a prism are often incorporated as part of an optical emitter and / or receiver. The imaging assembly **30**, connector **36** and/or imaging conduit **34** may be liquid-filled, such as with saline and may be flushed.

**[0065]** The imaging probe **12** may contain ports at one or more points along its length to facilitate flushing. For optical imaging, it is possible to consider a gas filled imaging probe **12**. Preferably, the gas would substantially comprise carbon dioxide or another readily dissolved gas. Alternatively, the imaging assembly may be compartmentalized such that there is at least one gas-filled compartment or lumen for optical imaging and at least one fluid-filled compartment or chamber for acoustic imaging.

**[0066]** The imaging conduit **34** comprises at least one optical waveguide and at least one conductive wire (preferably two or more) that connect an emitter and/or receiver via a connector to an adapter. The imaging conduit **34** may also act as a mechanical force transmission mechanism for rotating or translating the imaging assembly. For example, the imaging conduit **34** may comprise a fiber optic, wrapped by two layers of electrical wire that are insulated by each other. The imaging conduit **34** may further be reinforced by other structural features, such as helically wrapped wires or other designs used to construct imaging torque cables for rotating scan mechanisms, as described in the prior art.

**[0067]** The adapter **14** facilitates transmission of signals within any fibers and/or wires to the appropriate image processing units. The adapter **14** may also incorporate a pullback mechanism **49** (**Figure 2d**) or a reciprocating push-pull mechanism to facilitate longitudinal translation of the imaging assembly. Such longitudinal translation of the imaging assembly **30** may occur in con-

junction with the longitudinal translation of an external shaft that surrounds the imaging conduit **34**, or may occur within a relatively stationary external shaft.

**[0068]** Additional sensors may be incorporated as part of the adapter **14**, such as position sensing circuitry, for example to sense the angle of rotation of a rotary component within the imaging probe **12**. The imaging probe **12** may also include a memory component such as an EEPROM or other programmable memory device that includes information regarding the imaging probe to the rest of the imaging system. For example, it may include specifications regarding the identification of specifications of the imaging probe **12** and may also include calibration information regarding the probe **12**.

**[0069]** While precise alignment of the acoustic and optical imaging data is highly desired, it is also important to recognize the need to optimize the geometry of a minimally invasive probe so that it is as small as reasonably possible to achieve its desired purpose. Current IVUS probes are approximately 0.9 to 2 mm in diameter and the smaller sizes of probes can be delivered more distally within the vascular tree of the coronary anatomy as the vessel size tapers down. Thus, smaller sizes generally allow for interrogation of a larger portion of the coronary anatomy. It is therefore desirable to have embodiments of a probe that combines optical and acoustic imaging in arrangements that minimize certain dimensions of the probe, such as the diameter of the probe.

**[0070]** **Figure 2** is a perspective drawing of a flexible catheter containing a fiber optic **40** and a co-axial electrical wire **50**. The proximal connector contains fiber optic **40** that can be received by the adapter to optically couple the imaging fiber optic **40** to the optical imaging system "back-end". There are also electrical connectors **56** that allow the one or more electrical conduits to be connected to the ultrasound circuitry **20** and / or controller and processing units **18**. In embodiments where the imaging conduit rotates around its longitudinal axis, there may be a need to couple the rotating components of the imaging fiber optic with the relatively stationary fiber optic that connects to the optical imaging system's back-end **21**. The coupling of a rotating fiber optic probe can be accomplished using a fiber optic rotary joint incorporated either as part of the proximal connector of the imaging probe **10** or as part of the adapter **14**. Similarly, in embodiments where the imaging conduit rotates around its longitudinal axis, there may be a need to couple the conductive wires that rotate with the imaging conduit with the relatively stationary conductors of the ultrasound circuitry **20** and / or controller and processing units **18**, preferably by means of slip rings. These slip rings can be incorporated as part of the proximal connector of the imaging probe **36** or as part of the adapter **14**.

**[0071]** **Figure 2a** shows a cross sectional view of the mid section of the imaging probe of **Figure 2** taken along the dotted line which shows a fiber optic **40**, guidewire port **44** and guide wire **42**, imaging conduit **34**, imaging conduit lumen **46**, external sheath **48** which is a hollow,

flexible elongate shaft made of a physiologically compatible material and having a diameter suitable to permit insertion of the hollow elongate shaft into bodily lumens and cavities, and coaxial electrical wiring **50**. The expanded detailed view of the end of the imaging probe **10** shown in **Figure 2b** shows the distal end of the guidewire **42** extended beyond the end of the outer sheath **48** and a flush port **54** at the end of the sheath **48**. In **Figure 2** the proximal end of the imaging probe **10** includes another guidewire port **55** into which guidewire **42** is inserted and the connector assembly **36** which includes a flush port **58** and electrical contacts **56** along the connector body.

**[0072]** **Figure 2c** shows a schematic of how the rotary and non-rotary components of the imaging probe can be coupled with an adapter to the rest of an imaging system. **Figure 2d** schematically shows how the rotating components of the imaging probe can be coupled to the rotating components of an adapter. The rotating components of each can be electrically, optically and / or mechanically coupled using connectors and other configurations known in the art. Similarly, the non-rotating components of the imaging probe can be coupled to the non-rotating components of the adapter **14**. The adapter **14** can include slip rings, optical rotary joints and other such implements for electrically or optically coupling a rotary component to a non-rotary component and enable communication of necessary electrical and optical signals with the rest of the system.

**[0073]** Dual-fiber optical rotary joints are also available but considerably more complex. Electrical coupling between any conductor mounted onto a rotating component in the imaging probe **12** can be coupled to non-rotating conducting elements via metallic slip rings and springs, metallic slip rings and brushes or other commonly known methods of forming conductive contact between a stationary conductor and a rotary conductor.

**[0074]** While the electrical, optical and mechanical connections are shown separately in **Figure 2d**, it is possible to reduce the several connectors that must each be separately connected between the probe and adapter with fewer connectors by combining several connectors into combined connectors, as needed for a specific embodiment.

**[0075]** **Figure 3a** shows one embodiment of an over-the-wire configuration for an external sheath at **47** and **Figure 3b** shows a cross-section of sheath **47** through the portion that contains the imaging assembly **30** along the vertical line **3b-3b** in **Figure 3a**.

**[0076]** **Figure 3c** shows an embodiment at **60** that is a "rapid exchange" configuration for the external sheath that may be incorporated with the imaging probe if a guidewire is required. Sheath **60** in **Figure 3c** includes the entry port **55** shown in **Figure 2**. **Figure 3d** shows a cross-section of the "rapid-exchange" configuration **60** through the portion that is proximal to the entry port **55** for a guidewire along line **3d-3d** in **Figure 3c**. **Figure 3e** shows a cross-section along line **3e-3e** in **Figure 3c**.

[0077] The present invention describes several embodiments by which precisely registered ultrasound and optical images can be formed. The simplest conceptual approach is to have the paths of the ultrasound and optical imaging beams be aligned collinearly with each other.

[0078] Referring to **Figure 4a**, an imaging probe **399** is provided which is configured to allow imaging by acoustic and optical means in the same direction, so that an acoustic transducer that allows light energy to travel through a channel in the transducer is utilized. Essentially, probe **399** uses an acoustic transducer **402** that is altered to have an optically transmissive channel made through its substrate. The acoustic transducer **402** can be any kind of ultrasound transducer known in the art, such as piezoelectric composition (e.g. PZT or PVDF), a composite transducer or a single crystal transducer.

[0079] Electrical conductors **400** are directed to the conducting layers **401** on either side of the transducer's acoustic substrate **402**. A fiber optic **403** provides an optical conduit for enabling optical imaging. One or more matching layers can be added to the emission surfaces of the transducer, such as an epoxy layer (such as a silver or copper conductive epoxy layer which may functionally also serve as one or both of the electrodes that drives the transducer), or a polymer (such as parylene or PVDF).

[0080] The optically transmissive channel **407** is made by any of several techniques, such as precision drilling, laser ablation, photo-etching, inclusion of a feature in a mold to create the opening and others. Precision drilling may include the use of drill bits, such as diamond or carbide drill bits explicitly designed for cutting through hard materials. A high precision spindle, such as an air spindle, may be helpful for accurate and efficient execution of the drilling technique. A laser source can be used to ablate a channel through the substrate. Exemplary laser sources include YAG or excimer lasers.

[0081] Alternatively, if the acoustic transducer **402** is formed from a substrate that is initially viscous, a sacrificial component can be embedded in the piezoelectric during the formation of the piezoelectric transducer **402**. The sacrificial component can then be removed by mechanical means or exposure to a solvent. For example, a polystyrene cylinder can serve as the sacrificial component, which can be subsequently sacrificed using dissolution in acetone. Alternatively, if the piezoelectric material **402** is formed from a substrate that is initially viscous, a removable mandrel can be included in the material during the formation of the piezoelectric transducer and removed after the piezoelectric has partially or substantially hardened.

[0082] Conductive layers **401** on either side of the piezoelectric material **402** are incorporated as required for applying a voltage to the piezoelectric. The opening **407** is coupled to an optical waveguide **403**, either directly, or by means of one or more mirrors **404** or prisms **397** and one or more lenses **405**. If any optical components

are included within the opening, a dampening, insulating layer of a compliant material **406** (see **Figure 4l**), such as silicon or polymer may separate the optical components from the acoustic substrate **402** to act as either an electrical insulator or to minimize the transmission of stresses that are generated by the acoustic substrate **402** to the optical components.

[0083] As in **Figure 4b**, the light from the fiber can be directed towards a mirror **404** (or prism) that causes the light from the fiber to be deflected through the optically transmissive channel **407**. Alternatively, as in **Figure 4c**, a prism **397** can be used to deflect the light through the optically transmissive channel. The prism **397** may deflect light either as a result of total internal reflection or be assisted by a reflective coating on its deflecting surface **419**. The prism **397** may be a separate optical component that is affixed to the appropriate position along the optical path. For example, it can be glued in place onto the end of a fiber, onto a lens or onto a spacer using bonding methods such as UV cured glue. Alternatively, attaching a no-clad optical fiber along the optical path and cutting the segment of no-clad fiber at a desired length can be performed to make the prism. The segment of clad fiber can be cut and / or polished to achieve the desired angle. Mao describes this method in the previously cited reference.

[0084] Also seen in **Figure 4c**, an optically transparent window **409** may optionally be found at the end of the optically transmissive channel **407** and any unoccupied space within the channel may be filled with a gas, fluid or optically transparent material such as glass or any of several transparent polymers known in the art. The purpose of the window **409** is to prevent undesired air bubbles from being created or retained in the channel **407** and to protect the components in the optically transmissive channel **407**.

[0085] As seen in **Figure 4d** it may be desirable to have a gas instead of fluid or solid material inside the channel **407** to improve the refractive power of certain optical components such as a contoured lens **424**, which may be a ball lens.

[0086] As seen in **Figures 4e to 4g**, the GRIN lens **405** or other optical component can reside adjacent to the distal dip of the optical fiber **403**, between the fiber **403** and the deflecting mirror or prism **397** along the optical path. In this case, the opening **407** in the acoustic substrate **402** can be left free of any optical components and simply contain an optically transparent material, or be covered by a window **409**. Alternatively, the GRIN lens **405** or other optical component can reside in the optically transmissive channel **407** of the acoustic substrate **402**, as seen in **Figures 4g to 4l**. The sleeve of insulating material **406** mentioned above can surround the GRIN lens **405** or other optical component within the opening **407** as shown in **Figure 4l** in order to provide either mechanical or electrical insulation from the acoustic substrate **402**.

[0087] Referring to **Figure 4f** an optical spacer **433** is

located between the distal end of the optical fiber **403** and GRIN lens **405**. The optical spacer element **433** may comprise an optically transparent medium, such as no-clad fiber, glass, plastic, a gas-filled gap or a fluid-filled gap. The use of an optical spacer element **433** may help reduce the required precision for the alignment and sizes of optical components in order to achieve a desired focal length.

**[0088]** Alternatively, as seen in **Figure 4g**, the path length of the prism **397** or mirror can act as all or a portion of the optical spacer **433** in between the distal end of the optical fiber and the lens **405**. The advantage of using the distance that light must travel through the mirror or prism **397** as a substitute for a portion of a functional optical spacer is that the focusing element (e.g. the GRIN lens **405** or other lens) is closer to the region being imaged, thus improving the effective working distance of the optical imaging system. In some situations, the lens **405** can be offset from either edge of the optically transmissive channel to achieve the desired depth of focus, as in **Figure 4h**.

**[0089]** In other embodiments, it may be helpful to have one or more optical elements of the optical path extend beyond the outer surface of the acoustic transducer, such as element **434** as in **Figure 4i**, in order to achieve the desired performance of the optical imaging technique. This is particularly important when the acoustic transducer **402** is quite thin (such as a for very high ultrasound frequencies) or when the effective working distance of the optical imaging technique is longer than can be accommodated by having all the optical components reside below the emitting surface of the acoustic transducer.

**[0090]** It is also important to realize that the optical circuit can be distant from the surface of the acoustic transducer **402**. By way of example, as seen in the embodiment shown in **Figure 4j**, it may be desirable to have some backing material **435** interposed between the fiber optic **403** or other optical components proximal to the deflecting mirror or prism **397** and the acoustic transducer **402** to minimize back-reflections from the optical components.

**[0091]** The direction of propagation of the acoustic and optical imaging energy can be in a direction other than perpendicular to the longitudinal axis of the imaging probe. In fact, a slight angular offset of a few degrees is desired to minimize reflections back from the sheath that surrounds the probe. **Figure 4k** shows an embodiment of a probe that combines optical and acoustic imaging means aligned at an angle other than normal to the longitudinal axis of the probe.

**[0092]** The embodiment of the probe **500** shown in **Figure 5a** is structurally configured such that both acoustic and optical imaging sensors can be combined for viewing without components such as the mirror **404** of **Figure 4b** or prism **397** of **Figure 4c**. The head section of probe **500** containing piezoelectric material **402** for the acoustic sensor and the conductive layers **401** on either side of the piezoelectric material **402** is aligned along the longi-

tudinal axis of the fiber optic **403** and the probe is configured so that both acoustic and optical signals are emitted axially relative to the fiber axis, not perpendicular as in **Figure 4a**.

**[0093]** The embodiment shown in **Figure 5b** is analogous to the embodiment shown in **Figures 4b** and **4c**. **Figure 5c** is analogous to the embodiment shown in **Figure 4d**. The embodiment shown in **Figure 5d** is analogous to the embodiment shown in **Figure 4e**. The embodiment shown in **Figure 5e** is analogous to the embodiments shown in **Figures 4f** and **4g**. The embodiment shown in **Figure 5f** is analogous to the embodiment shown in **Figure 4i**.

**[0094]** **Figure 6a** shows the geometry of an emitting surface of a square transducer **402**. It should be noted that the geometry of the emitting surfaces of the acoustic transducers **402** are not limited to being in square in shape and may be any of several shapes, such as rectangular, circular, ellipsoid, and any other desirable shape. **Figure 6b** shows a square transducer with the hole **407** in the center, while **Figure 6c** shows a square transducer with a glass rod **501** in the hole **407**.

**[0095]** Results of a simulated beam profile using acoustic beam simulation software are shown in **Figures 6d** through **6f**, corresponding to the transducer geometries in **Figure 6a** through **6c** respectively. As can be seen, there is considerable similarity in the beam profiles of the various configurations, providing evidence that ultrasound transducers adapted to allow a channel for optical transmission are capable of producing an acceptable ultrasound beam profile suitable for imaging purposes.

**[0096]** A simpler method for aligning the optical and acoustic imaging means would be to place the fiber optic adjacent to the surface of the acoustic transducer **402** without going through the transducer **402** itself. **Figure 7A** shows an imaging probe **510** comprised of an acoustic transducer **402** with the distal end of an optical imaging circuit **428** placed on top of the acoustic transducer **402**. The distal end portion of the optical imaging circuit **428** comprises the distal end of fiber **403** and any optical components, such as an optical spacer **433**, a lens, such as a GRIN lens **405**, mirror **404** or prism **397**, that enable emission or collection of optical imaging energy. The distal end of an optical imaging circuit **428** can be affixed directly to the acoustic transducer **402** or supported by a support next to the acoustic transducer **402**. The distal end of optical imaging circuit **428** would affect acoustic signals generated and / or received by the acoustic transducer **402** as it lies directly in the path of a portion of the acoustic beam emitted by transducer **402**. However, a significant portion of the energy of the acoustic beam would not travel through the optical imaging means **403** and therefore would remain relatively unaffected.

**[0097]** Furthermore, the signal processing means preferably includes signal subtraction methods for discarding the portion of the signal that represents the early time portion of an echo signal to cancel reflections from inter-

faces close to the acoustic transducer's surface.

**[0098]** **Figure 7b** shows a perspective view of imaging probe **512** which is a modification of the system in **Figure 7a** where the distal end of optical imaging circuit **428** is recessed into the surface of the transducer **402** thus rendering the recessed portion of the transducer non-functional, so that acoustic beams transmitted or sensed by the acoustic transducer **402** do not substantially propagate through the overlying imaging fiber **403**. A top view of this embodiment is shown in **Figure 7c**. The portion of the transducer **402** rendered non-functional can be rendered non-functional by either removing the portion of the transducer **402** that lies underneath the distal end of optical imaging circuit **428** as shown in **Figure 7b**, or by electrically isolating the portion of the electrode underneath the optical imaging means. Removal may be done by several methods, including the use of a dicing saw to cut a channel through the transducer **402**. Furthermore, removal of a channel makes it possible to consider recessing the distal portion of the optical imaging means within a channel.

**[0099]** **Figure 7c** shows a top view of the emitting / receiving surface of the probe **510** shown in **Figure 7b** surrounding the distal end of optical imaging circuit **428**.

**[0100]** **Figure 7d** shows an imaging probe **516** that employs a composite transducer for the acoustic imaging means. In this case the composite transducer is a transducer comprising more than one signal generating element, or pillars **520**. The composite transducer in **Figure 7d** comprises four pillars **520**. The channels **522** in between the pillars **520** leave a channel **522** for one or more distal ends of optical imaging circuit **428** to be placed within the confines of the composite acoustic transducer. The distal end of an optical imaging circuit **428** need not necessarily be recessed within channels **522**, and can alternatively rest on or above the surface of the acoustic transducer **402**. Conducting connections **400** between the upper conducting surfaces of the pillars **520** allows for the pillars to be simultaneously activated. The channels **522** can be filled with a filler material, such as a polymer or epoxy, to increase the mechanical stability of the composite transducer, or to help affix the optical imaging means in place.

**[0101]** **Figure 7e** shows a top view of the imaging probe **516** with the distal end of the optical imaging circuit **428** placed within the center of the pillars **520**. Any of the implementations for the distal portion of the optical imaging circuit **428** (e.g. any combination of fiber optics, spacers, GRIN lenses, Ball lenses, air gaps, transparent windows), such as those shown in **Figure 4**, can be used in the implementations described in **Figures 7a to 7e**.

**[0102]** As part of most mechanical scanning mechanisms for imaging, there is a predominant motion associated with the scanning mechanism that defines the geometric path through which the imaging beam will sweep. For example, in an imaging system that uses a rotary motion to scan a region, there will typically be a circular or conical surface, through which the imaging beam

sweeps, with the circular or conical surface being centered approximately on the axis of rotation, as occurs in current implementations of mechanical scanning intravascular ultrasound. The predominant motion in this case is the rotational motion.

**[0103]** Alternatively, if the imaging emitter / receiver is translated along the longitudinal axis, then the imaging beam will sweep through a planar surface and the plane defined by that surface will include the axis of translation. This predominant motion in this case is a longitudinal translation.

**[0104]** If the imaging emitter / receiver is simultaneously rotated around a longitudinal axis of a probe and translated along a path that is generally parallel to the longitudinal axis of the probe, then the imaging beam will sweep through a surface defined by a helicoid geometry.

**[0105]** It is possible to generate co-registered images with good precision from multiple acoustic and / optical imaging means without having to have the two or more imaging beams be simultaneously collinear. This can be accomplished by having one or more imaging beams follow the path of a leading beam. Software or electronic circuitry can use knowledge of the speed and direction of the scanning mechanism's motions over time to then register the images generated from one of the imaging means onto one another.

**[0106]** For example, if the path of one imaging beam closely follows the path of another imaging beam (the leading beam) in a short time period, then it is possible to assume that the region scanned by the two means is similar enough to accurately co-register the two images with each other. The accuracy of the registration between the two images can be affected by the time delay in which the second beam follows the first beam. If the time delay is relatively small, then inaccuracies in the co-registration of the two images that could potentially develop in that time period are likely to be minimal. Such inaccuracies might include those caused by tissue motion (such as that induced by cardiac or respiratory motion), unintentional probe motion, physiologic changes such as blood flow and imprecision in the fidelity of the scanning mechanism. The time delay (which itself can vary over time) can be used for the process of registering the different images.

**[0107]** **Figure 8a** shows an example of an imaging assembly **530** that contains both an acoustic imaging means and an optical imaging means. The predominant scanning motion is a rotational motion around a longitudinal axis that lies along the length of the imaging probe. As illustrated, the acoustic imaging beam **532** and optical imaging beam **534** sweep through a path that is circular in nature. If the imaging beams are not aligned normal to the longitudinal axis, but rather at an angle other than 90 degrees from the longitudinal axis, then the path through which the imaging beams sweep will be conical in nature. If a longitudinal translation were to be applied in combination with the rotary motion, the two beams would follow a roughly helicoid path.

[0108] **Figure 8b** shows a side view of the combined imaging probe **530** where the acoustic beam **532** travels in one direction (upwards in the diagram) and the optical imaging beam **534** travels out of the page (towards the reader). In this case, the optical beam **534** and acoustic beam **532** at any instant are oriented 90 degrees apart from each other.

[0109] **Figures 8c through 8e** represent a time series of the rotational motion of the imaging probe **530** as it would appear from the distal end of the imaging probe. In this example, the optical imaging beam **534** leads the acoustic imaging beam **532** by 90 degrees of rotation. At a constant frame rate of 30 frames per second, the time delay that it would take for the trailing beam to become collinear with a prior position of the leading beam would under 9 milliseconds, which is a short period of time with respect to artifacts that might occur due to cardiac motion experienced by an intravascular catheter.

[0110] Given the importance of miniaturizing the space occupied by components and assemblies in minimally invasive imaging means, it may be desirable to recess some of the components. For example, as seen in **Figure 9a**, an imaging probe **540** has been configured to recess the distal end of the optical imaging circuit **428** into the backing **435** of the acoustic transducer **402**. Recessing may not only accomplish efficiency of space use, but it may also provide a method of fixing the distal end of an optical imaging circuit **428** to the acoustic transducer **402**.

[0111] The purpose of the backing material **435** on the acoustic transducer **402** is to attenuate signals generated from the back surface of the piezoelectric **402** so that an image is not formed by the energy that is emitted from the back surface of acoustic transducer **402** on which the optical emitter/receiver is located, but rather only from the primary emitting surface for acoustic signals (top surface) of the transducer **402**. Recessing an optical or other component in the backing material **435** may potentially cause the optical or other component to reflect signals back to the acoustic transducer **402** that would potentially create imaging artifacts.

[0112] **Figure 9b** shows a deflecting surface **544** in which the optical emitter/receiver is cradled that acts to deflect acoustic energy that might otherwise reach the optical emitter/receiver and deflects that energy laterally (substantially parallel to the surface of the acoustic transducer **402**) to minimize the amount of energy that is reflected back towards the transducer **402**. This deflecting surface **544** may be made of a hard substance such as glass or steel.

[0113] **Figure 9c** shows an implementation where the distal end of an optical imaging circuit **428** itself has a surface **545** that substantially deflects acoustic energy laterally without the need of an additional deflecting material as seen in **Figure 9b**.

[0114] For embodiments of imaging probes where the imaging beams scan as a result of rotational motion, it is not necessary that the rotational velocity be a constant or even remains in the same direction. It is possible to

have a reciprocating motion where the imaging assembly rotates in one direction and then stops and rotates in the opposite direction. In this situation, the leading and trailing beams swap roles with each other.

5 [0115] For example, in **Figure 10a**, the acoustic beam **532** initially follows the optical beam **534** as the imaging assembly rotates in a counter clockwise direction. The acoustic beam **532** continues to follow the sweep path of the optical beam **534** as shown in **Figure 10b** until the rotational velocity of the imaging probe reaches zero, (as in **Figure 10c**). Once the direction of rotation changes to the opposite direction, the acoustic beam **532** becomes the leading beam and the optical beam follows (as in **Figures 10d and 10e**). The motion can change direction as many times as desired with a concomitant change in the definition of the leading and trailing sensor beams.

10 [0116] **Figure 11** shows an imaging probe **540** where the predominant motion is a longitudinal motion back and forth along arrow **541** where the surface swept the optical beam **534** and the acoustic beam **532** are two co-planar rectangles. As the imaging assembly is translated proximally (to the left in **Figure 11**) the optical imaging beam **534** leads the acoustic imaging beam **532**. The opposite is true for distal translation (to the right in **Figure 11**). The longitudinal motion can be reciprocated as well.

15 [0117] With either longitudinal or rotational predominant motions, it is understood that additional motions can be combined with the predominant motion. For example, a slow translation (such as 10 mm / s or less, and typically 1 mm/s or less) can be added to a rapid rotational scanning motion (such as 360 degrees per second or more and typically 3600 degrees per second or more) in order to acquire 2D cross-sectional images at different longitudinal positions.

20 [0118] Similarly, a slow rotational motion (e.g. less than 360 degrees per second and typically less than 30 degrees per second) can be added to a sequence of rapidly reciprocating longitudinal motions (averaging over 0.1 mm/s and more typically more than 1 mm/s) to create a series of longitudinal images acquired at different orientations around the longitudinal axis of the imaging probe. The alignment of the various imaging elements at the distal end is configured such that the one of the imaging beams will follow the other during the predominant motion, but the ability to accurately register the images on top of each other would not be significantly affected by the addition of a relative slow secondary motion. While absolute numbers for slow and rapid motions in the rotational and translation motions are provided above, it is the relative magnitude of these motions that is more important.

25 [0119] Collinear alignment of the optical and acoustic beams (as shown in the embodiments shown from **Figures 4a to 5f**) provide very accurate registration of the optical and acoustic images. An alternative embodiment of the probe is configured to have the optical and acoustic beams substantially overlap each other by angling either the optical imaging emitters / receivers towards the path

of the acoustic beam or by angling the acoustic imaging emitter towards the path of the optical imaging beam. **Figure 12** shows such an embodiment of an imaging probe **546** where the distal end of an optical imaging circuit **428** is configured such that the optical imaging beam **534** is angled towards the acoustic imaging beam **532** and vice versa. This provides a simpler method of construction than aligning the optical and imaging beams as seen in **Figures 4a to 5f**, but allows the two imaging means to provide what may be a reasonably precise overlap over a portion of the two imaging beams. In particular, embodiments whereby the beams are aligned such that they overlap over a substantial portion of their focal ranges would be useful.

**[0120]** **Figure 13** shows an embodiment of the imaging probe **550** configured to image simultaneously in the same general orientation and from the same general origin with both acoustic and optical means. At least one fiber optic **410** and one electrical conduit **411**, such as a pair of coaxial conductors, reside within the imaging conduit **560** and travel to the imaging assembly **562**. The imaging assembly **562** comprises an acoustic transducer **412** configured for imaging in a substantially side-viewing direction indicated by arrow **420**. The imaging assembly **562** also includes distal end of an optical imaging circuit **564** configured for imaging in a substantially side-viewing direction indicated by arrow **421**.

**[0121]** The acoustic transducer **412** and distal end of an optical imaging circuit **564** are configured such that they allow imaging in two or more separate directions at any instant within the same cross-sectional plane that is substantially perpendicular to the axis **423** around which the imaging assembly **562** rotates. Thus, assuming minimal translation of the imaging assembly **562** while the imaging assembly is rotated, the imaging data collected by the optical emitters/ receivers **564** can be co-registered with the imaging data collected by the acoustic transducer **412**. For example, if the acoustic and optical means are configured to image in directions that are 180 degrees opposite of each other around the longitudinal axis, as shown in **Figure 13**, then the region imaged by the acoustic transducer **412** at one point in time will be substantially the same region that is imaged by the distal end of an optical imaging circuit **564** after the imaging assembly **562** has been rotated by half a revolution. Similarly, if the imaging beams **420** and **421** have a similar angle from the longitudinal axis other than 180 degrees, they will both sweep through paths of substantially coincident cones, and can therefore be co-registered.

**[0122]** The embodiment of the probe **570** shown in **Figures 14a and 14b** is configured such that both IVUS and OCT can be combined for forward viewing with a deformable component. At least one fiber optic **410** and one electrical conduit **411**, such as a pair of coaxial conductors reside within the imaging conduit **578** and travels to the imaging assembly **572**. The acoustic transducer **412** is configured for imaging in a substantially forward-looking direction indicated by arrow **413**. A distal end of an

optical imaging circuit **574** is configured for imaging in a substantially forward-looking direction indicated by arrow **414**.

**[0123]** The distal end of an optical imaging circuit **574** typically comprises a distal end of a fiber optic **410** combined with a lens **415**, such as a GRIN lens and an optional spacer (not shown). The imaging conduit **578** comprises an artificial muscle actuator that has the property of being able to deform upon the application of an electrical charge. **Figure 14b** illustrates how the imaging angle would be changed if an artificial muscle actuator achieved a deformation while **Figure 14a** shows the shape of the probe without application of a voltage to actuator.

**[0124]** Embodiments of the present imaging probe may be configured to make use of a deflector to allow for a larger transducer to be used within the imaging probe. Alternatively, the deflector may be pivotable and coupled to a pivoting mechanism to enable an additional degree of freedom in the scanning mechanism. For example, the scanning mechanism may facilitate 2D imaging, or may augment a 2D imaging system into a 3D imaging system. Alternatively, the deflector may be translated along the longitudinal axis in order to change the focal depth of the imaging system.

**[0125]** **Figure 15a** illustrates an embodiment of an imaging assembly **590** that comprises a deflector **592** used to deflect optical and / or acoustic imaging energy into a generally radial direction. The deflector **592** is made of one or more reflective materials. Optically reflective materials include polished or sputtered metals, such as stainless steel, gold, silver and platinum.

**[0126]** Acoustically reflective materials include stainless steel and other metals, quartz and other crystals, glass and hard polymers. **Figure 15b** shows another embodiment of an imaging assembly **600** which comprises a deflector **602** that pivots around a pivot point **604** and thus allows the angle between the imaging beam and the longitudinal axis of the imaging probe to vary. The imaging assembly **600** may be configured so that deflector **602** can change position by being coupled to a variety of mechanisms, including mechanisms which utilize centripetal motion, magnetic forces, cable mechanisms, rheologic forces, piezoelectric drivers, miniaturized motors and others.

**[0127]** **Figure 15c** illustrates an embodiment of the arrangement in **Figure 15b** wherein a cantilever **901** mounted on a cantilever mount **902** and the deflector's range of motion is limited by a minimum stop **82** and a maximum stop **80**. This embodiment has the property of having the imaging angle change as a result of changes in the rotational motion of the imaging assembly around the longitudinal axis of the probe. At rest or low rotational speeds, the cantilever wire forces the deflector **602** around its pivot point such that it comes into contact with stop **80**. At higher rotational speeds, centripetal acceleration causes the deflector **604** to pivot away from stop **80**. As centripetal acceleration continues to overpower

the restoring force exerted by cantilever **901** on deflector **602**, the deflector eventually comes into contact with stop **82**. In such an embodiment, an imaging assembly **600** with a 3D scanning mechanism is implemented.

**[0128]** Figure **16a** illustrates an embodiment of the distal portion of an imaging probe **100** capable of both acoustic and optical imaging in a generally forward-looking direction. Figure **16a** shows an embodiment of a distal end **29** of an imaging probe containing an imaging assembly **30** that includes a tiltable component **70** where the tiltable component is a disc mounted on a pivoting mechanism such as a pin **72** that extends through the disc **70**. The pivoting mechanism **72** defines the tilting axis of the tiltable disc **70**. When the imaging assembly **30** is at rest, the disc **70** will remain in an arbitrary starting position. However, as the imaging assembly **30** rotates, the disc **70** will align itself such that the normal of the planes defined by the faces of the disc **70** are substantially parallel with the longitudinal axis **75**. The disc **70** has two preferred orientations when the imaging assembly **30** is rotated, that are separated by a rotation around the tilting axis of 180 degrees.

**[0129]** For the purposes of this description, the tilt angle will be referred to as the angle between the longitudinal axis **75** and an imaginary axis through the tiltable component **70** that is parallel to the longitudinal **75** axis when the tiltable component **70** is in one of its preferred orientations. By way of example, when the tiltable component **70** is in a preferred orientation, the tilt angle is approximately zero. If the tiltable component **70** is tilted away from its preferred orientation by an external force, such as gravity, magnetic forces, electrostatic forces, friction with another moving part or fluid, compressive forces, normal forces or any other source of incompletely opposed torque on the tiltable component **70** around the tilt axis, the tilt angle will increase.

**[0130]** One or more mechanisms may be included in the imaging assembly **30** that tends to cause the tiltable component **70** to have its tilting angle increase. For the purposes of this invention, such a mechanism is referred to as a restoring mechanism. A torsion spring **76** (as shown in Figures **16a** and **16c**), a cantilever or a compression spring can be used as a restoring mechanism, where one end of the spring **76** is mechanically in contact with tiltable component **70** and the other end is mechanically in contact with another part of the imaging probe **100**, such as the body of the imaging assembly **30**.

**[0131]** Alternatively, magnetic, electrostatic, hydraulic or other mechanisms that apply a torque on the tiltable component around the tilting axis could be applied. Other examples of mechanisms that could be used to provide a restoring force include tension from an elastomer (such as rubber, polyurethane, silicone, fluoroelastomers, thermoplastics and many others) or by use of a cantilever spring or foil, such as springs or foils made of platinum, nitinol, steel or other suitable materials. In very small embodiments of the imaging device, where intermolecular forces such as electrostatic forces and Van der Waals

forces between components in the imaging assembly may become quite significant even without the application of an external voltage. Therefore, the innate intermolecular forces between the tiltable component and structures close to the tiltable component, such as the stops **80** and **82** described below, may be sufficient to provide a net restoring force. For example, a stop comprising a surface made of PVC, nylon or LDPE could provide sufficient attraction between the tiltable component and the stop.

**[0132]** One or more stops **80** and **82** may limit the range of the tilt angle of the tiltable component **70**. For example, a post or lip **80** can extend from the shell **84** of the imaging assembly **30** as a stop to prevent the tilting component from further changing its tilt angle while it makes contact with the stop **80**. Therefore, a stop can be used to limit the tilt angle from exceeding a maximum value determined by the position of the stop. In many embodiments, this maximum tilt angle is the tilt angle that is achieved when the imaging assembly **30** is at rest and at low rotational speeds.

**[0133]** An additional or alternative stop **82** can be included to create a minimum tilt angle that the tiltable component will achieve at rotational speeds in the upper end of the operating range. Indeed, there are many situations in which there is no significant benefit in allowing the tilt angle to reach zero, as will become apparent in the following descriptions of specific embodiments. Figure **16c** shows the tiltable component hitting the second stop to limit its range of motion at higher rotational speeds of the imaging assembly.

**[0134]** The imaging assembly may include both optical emitters and associated optics and ultrasound transducers. The ultrasound transducer **88** is mounted at the end of small coaxial cable **89** and lens **92** and mirror **94** are mounted at the end of a fiber optic cable **96** in the imaging assembly **30** in Figures **16a** to **16d** with the optical and ultrasonic emitters configured to focus imaging energy onto the tiltable component **70**. The ultrasound transducer **88** and optical emitter can direct imaging energy towards the tiltable component **70**. Alternatively, one of the embodiments that enables collinear optical and acoustic imaging, as seen in Figures **4a** through **4k** or Figures **5a** through **5f** can direct imaging energy towards the tiltable component **70**.

**[0135]** The imaging energy is then deflected by an energy-deflecting component mounted on the tiltable component **70**. For ultrasound imaging, the energy-deflecting component (the tiltable component **70**) may comprise an acoustically reflective surface, such as a solid metal surface (e.g. stainless steel) or crystalline surface, such as quartz crystal or glass. For optical imaging, the energy deflecting component (tiltable component **70**) can comprise an optically reflective surface such as a mirror surface made from polished metal, metallized polymer such as metallized biaxially oriented polyethylene terephthalate (Mylar), sputtered or electrochemically deposited metal or metal foil. Metals commonly used to make mir-



rors include aluminum, silver, steel, gold or chrome.

**[0136]** Alternatively, the energy-deflecting component could be made of a transparent refractive material, such as glass, clear polymers, and many others, and deflect the imaging energy in a manner similar to a prism. Preferably, the emitter and / or receiver is mounted on a component of the imaging assembly that rotates with the imaging assembly. However, it is also possible that the emitter and / or receiver is mounted on a component of the imaging probe that does not rotate with the imaging assembly while the energy deflecting mechanism within the imaging assembly does rotate. This could be achieved by mounting the emitter and / or receiver on an external sheath for example, or by having the imaging assembly divided into two or more sub-assemblies, one of which rotates and includes the tiltable component.

**[0137]** For ultrasound and optical coherence tomography, the ability to adjust the angle of propagation of the emitted and/or received imaging energy, when combined with the rotational motion of the imaging assembly, allows a 3D volume to be scanned. For angiography and infrared imaging, the ability to adjust the angle of propagation of the emitted and/or received imaging energy, when combined with the rotational motion of the imaging assembly, allows an image to be produced using a single fiber optic rather than requiring a bundle of fibers. Such an improvement can result in greater flexibility and / or miniaturization of the imaging device.

**[0138]** In the case where the energy-deflecting component comprises a reflective surface it is not necessary that the reflective surface be planar. For example, in the case of acoustic imaging, it may be advantageous for an acoustically reflective surface to have a contour to it, such as a parabolic or spheroid contour, so that the acoustic beam can be focused by the reflective surface and improve lateral resolution of the acoustic imaging system as a result. Furthermore, in the case where the tilting component is used to deflect both acoustic and optical energy using reflection, the acoustic reflector need not be the same surface that reflects the optical energy.

**[0139]** For example, while it might be advantageous to have a contour such as a parabolic contour for the acoustically reflective surface, it may be preferable to have a planar surface for the redirection of the optical imaging energy. This can be accomplished by having an acoustically reflective surface such as a stainless steel disc with one of its faces contoured to have a parabolic shape to it as in **Figures 17a** through **17d** which show a tiltable deflecting component that has an optically reflective surface that is distinct from the acoustically reflective surface.

**[0140]** **Figure 17a** is a perspective drawing of a deflector that has holes on its side for receiving pins on which the deflector can pivot within an imaging assembly. **Figure 17b** shows a cross-section through the deflector near the center of the deflector. The holes for receiving pins **465** are seen. The top layer is a flat, optically reflective layer **461**. Under the optically reflective layer **461**

a generally acoustically transparent layer **462**, which lies between the optically reflective layer **461** and an acoustically reflective substrate **463**. **Figures 17c** and **17d** show cross-sectional images of such a deflector at different points away from the center of the disc.

**[0141]** Such a deflector can be constructed by taking a disc of an acoustically reflective material such as stainless steel and drilling the necessary holes or indentations so that the deflector can eventually be mounted into an imaging assembly. A parabolic or spheroid indentation can be made into one face of the disc. The indented surface can then be filled with an acoustically transparent medium, such as polymethylpentene (TPX). A thin layer of gold, silver or chrome can be sputter deposited onto the exposed planar polymer surface to act as an optically reflective surface. Such a layer may be on the order of 300 Angstroms to 20,000 Angstroms such that it is thin enough that its mechanical properties to allow acoustic energy to transmit through it, while simultaneously providing an optically reflective surface.

**[0142]** The result of such a fabrication process is to create a layered reflector that reflects acoustic energy from the contoured surface to achieve the desired focusing effect, while the optical energy is reflected from a planar surface. It is a further advantage of this construct that the optical and acoustic imaging can occur in a configuration where the optical and acoustic imaging energy travels through the same general space, facilitating co-registration of optical and acoustic images and minimizing the amount of space required within the imaging assembly to accommodate more than one modality of imaging.

**[0143]** In some embodiments, such as the assembly shown in **Figures 16a** and **16c**, it may be helpful to use one of the imaging modalities solely to measure a parameter useful for the reconstruction of 2D and 3D images. For example, in the case of a volumetric imaging probe that uses a deflectable component, it may be desirable to use OCT to accurately measure the tilt angle of the deflectable component. Thus, an ultrasound image could be generated with knowledge of the tilt angle derived from OCT data, such as the tilt angle of tiltable component **70** in **Figure 16a** without necessarily using the OCT data to generate corresponding OCT images of the region outside of the imaging probe.

**[0144]** In some embodiments, it will be desirable to have more than one method for optical imaging in an intravascular imaging system. For example, OCT and angiography may be a useful combination. **Figure 18a** shows an ultrasound imaging transducer **402** with two (2) distal ends of optical imaging circuits **428** through two (2) separate optically transmissive channels in the acoustic transducer. **Figures 18b** and **18c** show an acoustic imaging transducer with two (2) distal ends of optical imaging circuits **428** arranged in a manner such that they are aligned along the predominant rotary motion of the imaging assembly. These are examples of using more than one optical imaging emitter / receiver at the distal

end of the imaging probe. If the imaging probe uses extensive rotary motion around its longitudinal axis as part of the scanning mechanism, such embodiments may require the use of a multi-channel optical rotary joint.

**[0145]** Alternatively, the optical imaging light sources and / or detectors for some of the imaging systems may be mounted on the rotary portion of the imaging probe and be coupled to the imaging system using electrical slip rings or wireless communication. A battery may optionally be used as a source of electrical energy on the rotary portion of the probe or adapter to minimize the number of slip rings required. Illuminating sources and photodetectors can be placed at the proximal end of the imaging probe and may be configured such that they rotate around the longitudinal axis of the probe with the rest of the imaging conduit **34** so that further optical couplers are not required between the imaging probe and the adapter. This is done because the complexity of rotary optical joints increases substantially if more than one fiber is involved to connect the probe to the rest of the system.

**[0146]** If the imaging probe uses only reciprocal rotary motion over a short range of angles (such as less than two full revolutions), or no rotary motion at all, then the use of an optical rotary joint is not necessary, simplifying the task of coupling the optical elements of the imaging probe to the image processing and display hardware.

**[0147]** In another embodiment, it is possible to using the same optical imaging emitter / receiver at the distal end of the imaging probe and use optical routing circuitry such as switches, multiplexers, demultiplexers, prisms, diffraction gratings, couplers and / or circulators to use the same fiber and distal optical components for more than one imaging modality. **Figure 19** shows a schematic of a system where there are two (2) optical imaging systems **211** that are coupled to the same optical imaging waveguide **212** via optical routing circuitry (comprising one or more of the components listed above). The waveguide may be coupled to the imaging probe via an optical rotary joint **213** if the image probe **12** requires a large range of rotary motion as part of its scanning mechanism. The distal end of optical imaging circuit **428** may comprise any of the combinations of optical fiber, spacers, mirrors, prisms, ball lenses, GRIN lenses, air gaps and transparent windows mentioned elsewhere in the present invention to enable optical imaging. While many optical imaging elements, such as the waveguide and lenses, are designed to operate optimally for particular ranges of wavelengths (e.g. infrared vs visible spectrum), the performance of a fiber optic or other optical component designed for one range is often still adequate to provide information using light in the other spectrum.

**[0148]** Therefore, imaging using more than one range of wavelengths can occur simultaneously. Alternatively, the imaging waveguide can be used at different time intervals for different imaging modalities by means of optical switches, multiplexers and demultiplexers within the optical routing circuitry **210**, or by simply timing the use

of the optical waveguide at different time intervals for different imaging modalities.

**[0149]** While a fiber optic would be a preferred optical waveguide **212** for most embodiments, it may be desirable to use an alternative form of optical waveguide that is potentially more space efficient than an optical fiber. For example, a thin optical channel, on the order of 3 to 500 microns in maximal diameter and preferably on the order of 4 to 125 microns can be formed in a catheter at the time of extrusion. A fluid medium with a high index of refraction can be introduced into the optical channel, such as by means of injection. Such a fluid medium may include an epoxy or adhesive specifically designed for optical components.

**[0150]** The fluid medium may also be curable, such as in the case of UV curable adhesives. The creation of an optically transparent channel filled with a material with a high index of refraction surrounded by the extruded catheter material with a lower index of refraction would essentially replicate the functionality of including a fiber optic, but may allow for slightly more efficient use of space in the catheter by not requiring a separate cladding layer. The optimal use of space in a catheter is often important given their minimally invasive nature and the limited space available in the regions in which these catheters are deployed.

**[0151]** Yet another mode of operation for the present invention is the use of a transducer that combines acoustic transduction with an optical transducer where the transmitted energy is of one form and the received energy is of another. For example, photoacoustic imaging comprises delivery of light-based energy to an imaged region. The photons interact with the imaged region and create acoustic energy as part of their interaction with the medium in which they propagate. This acoustic energy is often in the form of ultrasound waves, and can be detected by an ultrasound transducer. It should be apparent that the use of an optical emitter aligned and in combination with an acoustic receiver would be a good configuration to enable photoacoustic imaging. An ultrasound transducer with an opening for optical imaging or that allows substantial overlap in the acoustic and optical imaging regions, such as those shown in **Figures 4a** through **4k**, **5a** through **5F** or **Figure 12**, would enable photoacoustic imaging.

**[0152]** Similarly, sonoluminescent imaging comprises delivery of ultrasound-based energy to an imaged region (Daniels and Price, *Ultrasound in Medicine and Biology* 1991:17(3):297-308). The acoustic energy interacts with the imaged region and creates photons as part of its interaction with the medium in which it propagates. Some of these photons are directed back toward the source of the acoustic energy. It should be apparent that the use of an ultrasound transducer aligned in combination with an optical receiver would be a good configuration to enable sonoluminescent imaging. I

implementations of acoustic and optical imaging elements where the imaging beams are collinear, or sub-

stantially overlap, such as those shown in **Figures 4a** through **4k**, **5a** through **5f** or **Figure 12**, would enable sonoluminescent imaging.

**[0153]** Referring to **Figure 1** again, imaging probe **12** (which may include any of the embodiments of the acoustic and optical sensors discussed herein) and its components may be of several dimensions and properties depending on the anatomic location and purpose of use for the imaging that is enabled by the imaging probe **12**. For example, for the purposes of use in the cardiovascular system, including the cardiac chambers, the imaging probe **12** would preferably be elongate and flexible, with a length ranging from 5 to 3000 mm, preferably with a length ranging from 300 mm to 1600 mm. The imaging conduit **34** and imaging assembly **30** may have a maximum cross-sectional dimension ranging from 200 microns to 10 mm, preferably ranging from 500 microns to 5 mm. An external sheath **48** may surround both the imaging conduit **34** and imaging assembly **30**. This would enable the imaging conduit **34** and imaging assembly **30** to rotate within the external sheath while mechanically isolating the rotational motion of these two components from the surrounding tissues.

**[0154]** In yet another example, the use of the imaging probe **10** in the gastrointestinal system would typically have the imaging probe **10** being elongate and flexible, with a length ranging from 100 mm to 2000 mm and preferably in the range of 300 mm to 1500 mm. The maximum cross-sectional dimension would typically range from 3 mm to 20 mm.

**[0155]** In yet another example, the use of the imaging probe **10** to image soft tissue via percutaneous means would have the imaging probe with a rigid shaft. The external sheath would be replaced by a rigid hollow shaft, such as a stainless steel tube although many other polymers, metals and even ceramics would be functionally suitable.

**[0156]** In yet another example, the use of the imaging probe **10** in the intraoperative neurosurgical setting would typically have the imaging probe **10** being short and semi-flexible, with a length ranging from 50 mm to 200 mm. It would be preferable that the surgeon can bend and shape the probe during the procedure to provide optimal passage from extra-cranial space towards the intracranial target being imaged. The maximum cross-sectional dimension would range from 200 microns to 5 mm and preferably from 500 microns to 3 mm.

**[0157]** In yet another example, the use of the imaging probe **10** in the interventional neurovascular setting would typically have the imaging probe **10** being long and ultraflexible, with a length ranging from 200 mm to 4000mm and preferably ranging from 1300 mm to 2000 mm. The maximum cross-sectional dimension would range from 200 microns to 5 mm and preferably from 500 microns to 3 mm. The distal end of the probe would preferably possess shape memory to enhance navigation through the neurovasculature.

**[0158]** Embodiments of the present invention can be

used in conjunction with or incorporated into devices that are used for intervention, such as those used for cardiovascular intervention, such as an angioplasty balloon, atherectomy device, stent delivery system or localized drug delivery system. It can also be used in conjunction with or incorporated into devices that facilitate biopsies, radio-frequency ablation, resection, cautery, localized brachytherapy, cryotherapy, laser ablation or acoustic ablation.

**[0159]** In particular, using the image scanning mechanism to direct higher powers of optical or acoustic energy to a targeted region can facilitate the use of the current device to enable laser or acoustic ablation of tissue. For example, while imaging a region of a blood vessel with an OCT or ultrasound embodiment of an imaging probe described in the present invention a region for the delivery of therapy can be selected through a user interface. Then, powerful pulses of energy can be delivered at times when the scanning mechanism is oriented to delivery energy in the desired direction. For example, pulses of laser energy can be transmitted down the same fiber optic used for optical imaging, be deflected by a deflecting component in those embodiments that include a deflecting component, and travel towards the targeted tissue for the desired effect. The timing of the pulses of laser energy is coordinated with the scanning pattern realized by the imaging probe to direct the energy towards the targeted region.

**[0160]** The opportunity to acquire accurately registered images of two or more high resolution imaging modalities provides significant information that is likely to be more useful than available by a single imaging modality. Maschke et al describe the formation of a composite image whereby the inner portion of an intravascular image is composed of OCT imaging information while the outer portion of an intravascular image is composed of IVUS imaging information. This takes advantage of the higher resolution images acquired by OCT and the higher penetration of IVUS. However, the reliability of this superposition of IVUS and OCT images is limited by the inaccuracy of the registration in the IVUS and OCT images that occurs using the arrangement of the IVUS and OCT imaging elements as described by Maschke and are substantially overcome by many of the embodiments in the present invention.

**[0161]** Alternative presentations of combined IVUS and OCT images might include dividing the image into sectors, where alternating sectors are displayed using alternating imaging means, as seen in **Figure 20a**. First image **231** and second image **232**, where the first and second images are co-registered with each other images and acquired by different means, can be used to form a combined image **234** where sectors **233** of the first image replace sectors of the second image. Optionally, the borders **235** defining the sectors **233** can rotate over time around the center of the image to provide a dynamic image for identifying features in both the first and second co-registered images. **Figure 20b** shows a time progres-

sion of the rotations of the borders **235** around the center of the combined image **234**.

**[0162]** Alternatively, the user can specify which portions they would like to have as one image and which they would like to see as the other by identifying closed contours **236** in the second image as seen in **Figure 21 a** or by identifying a space **237** in between two closed contours in the second image, as seen in **Figure 21 b**.

**[0163]** Alternatively, displaying the first image **231** and second image **232** at the same position on the screen as separate layers and varying the transparency of the layer in the foreground can effectively provide a means for combining the images. Alternatively, the order of the layers can be varied over time, such as by having the IVUS image in the foreground for one time interval and then transitioning to having the OCT image in the foreground for a subsequent time interval, as seen in **Figure 22**.

**[0164]** It is an object of the present invention to be able to identify certain features of interest in a first image **231** and transfer knowledge of that feature (such as its position, shape, signal properties or composition) to a second image **232** that is accurately co-registered with the first image **231**. Geometric features include specific points, contours or 2D regions in an image. As seen in **Figure 23a**, a user can identify a point **238**, contour or region in a first image **231** manually, through the user interface of the imaging system (such as with a mouse or keyboard) and have that geometric point **238** appear in a second image **232** co-registered with the first image **231** as in **Figure 23b**. The availability of one or more other images that are accurately co-registered with the first image makes it possible to superimpose any or all of the geometric features from the first image to any of the other images.

**[0165]** By way of example, the user might identify the inner boundary of a blood vessel or the trailing edge of a fibrous cap in an OCT image. **Figure 24a** shows the contour representing the inner border **241** identified in a schematic representation of an OCT image (the first image). Similarly, the outer boundary **242** of the vessel wall (usually defined by the external elastic lamina) can be identified in an IVUS image (the second image). The contours representing the inner boundary **241** of the blood vessel or the trailing edge of the fibrous cap can then be superimposed onto the corresponding IVUS image. Similarly, the outer boundary **242** of the vessel wall (usually defined by the external elastic lamina) can be identified in an IVUS image. The contour representing the outer boundary as assessed in the IVUS image can be superimposed onto the OCT image. **Figure 24b** shows the inner and outer boundaries on both the first and second images.

**[0166]** While the inner boundary of the blood vessel is readily identified on most IVUS images, the OCT generated contour would be more accurate in most circumstances. Furthermore, OCT is thought to be much better for identifying the fibrous cap of a plaque, in part due to its higher resolution. However, IVUS can see much fur-

ther into most vascular tissues and can provide a better assessment of the outer vessel wall.

**[0167]** A geometric feature can include features observed in 3D data sets, such as surfaces or volumes. A surface or volume observed in a 3D imaging dataset can be superimposed into another 3D imaging dataset if the two imaging datasets are accurately registered.

**[0168]** The geometric features of interest need not be manually identified. It is possible that features in an imaging dataset can be identified by automated or semi-automated means to minimize user intervention. For example, there are several border detection methods cited in the literature on IVUS (e.g. Klingensmith, IEEE Transactions on Medical Imaging, 2000;19:652-662). Automated border detection methods analyze an image to identify a contour of some pre-determined significance. Semi-automated methods are similar, but require some user intervention to either provide a starting point for the border detection algorithm or to refine the results produced the algorithm.

**[0169]** Other feature detection algorithms can be conceived of to identify features other than a border. For example, a hyper-intense / bright region in an ultrasound image followed a dark region in the same direction of the imaging beam is often referred to as "shadowing" and occurs most commonly when the area being imaged includes either calcium (such as from advanced atherosclerosis or malignant processes) or metal (such as from stents or other implants). Similarly, a highly intense region in an OCT image of a blood vessel, followed by a rapid but continuous attenuation of the signal acquired along the same imaging path is suggestive of necrotic material in the vessel wall. It is possible to detect such regions algorithmically and identify them in their respective images. Once such features are identified in their respective images, their position and shape can be superimposed into other images that are accurately co-registered.

**[0170]** In certain embodiments of the present invention, it will be desirable to do some adjustment to one or more of the images to further improve the co-registration. While many of the embodiments of the present invention improve the precision of acquiring imaging data with one or more imaging methods, there may be some advantage to further adjusting the images to improve the accuracy of the co-registration process. For example, ultrasound images are generated assuming a constant speed of sound through all tissues, while OCT assumes a constant speed of light through all tissues.

**[0171]** In reality however, there are small changes in these speeds depending on the composition of the tissue in which each of the imaging energies propagate. Therefore, prior to completing the co-registration process for one or more images, it may be desirable to morph or warp one or more of the images by identifying certain features in the two or more images that are to be co-registered and using those features to guide the morphing process. Any point, contour or other feature iden-

tified in all of the images to be co-registered can be used to drive the morphing process. An ultrasound image is most commonly formed by displaying a grayscale representation of the intensity of the ultrasound signal reflected back from the approximate anatomic location that corresponds to each pixel in the image. Similarly, an OCT image is most commonly formed by displaying a grayscale representation of the intensity of the light reflected back from the approximate anatomic location that corresponds to each pixel in the image.

**[0172]** Aside from the intensity information at each location in either an ultrasound or OCT image, there are several other features from ultrasound or OCT images that can be very helpful for analysis derived from combined imaging.

**[0173]** The display of an image derived from ultrasound signals based on a feature other than then intensity of a sample in the image is well known in the art. Nair et al (Circulation 2002; 106(17):2200-2206 and US patent number 6,200,268) published results of an algorithm that measures several parameters of an ultrasound signal in discrete regions of IVUS images of blood vessels. Each region was also assigned a tissue category based on histological analysis of the vessel. The ultrasound derived parameters and the histological classification of each region were input into a pattern recognition engine to generate an algorithm that is subsequently applied in an attempt to classify tissue in vivo based on its many ultrasound signal properties. Some of the properties used for analysis include frequency domain parameters over a defined range of frequencies such as maximum power, frequency of maximum power, minimum power, frequency of minimum power, slope, y-intercept, mid-band fit and integrated backscatter. The image generated comprises a topographical map of the vessel cross-section and a discrete number of colors, with each color representing a single tissue category. Wilson et al demonstrated the use of measuring the frequency domain attenuation of an ultrasound signal in regions of an IVUS images and overlaying a color map of the attenuation slope onto the conventional IVUS image to identify areas thought to correspond to specific pathological types.

**[0174]** Similarly, features of interest can be measured or identified in optical images in order to generate images other than intensity-based images. Parameters or other features that can be used to generate such images include attenuation, polarization sensitivity, detected edges, spectroscopic information and others.

**[0175]** As a result of the high degree of accuracy of co-registration enabled by the present invention, it is possible to generate images based on features or signal properties measured with more than one imaging modality. For example, a composite image can be made using an inner border **245** identified by OCT, an outer border **246** identified by IVUS and a color map of the most likely tissue components within the vessel wall using a pattern recognition system that combines optical signal properties with acoustic signal properties within focal regions

of the imaging datasets to generate a composite image that will improve the ability to identify important components within the vessel wall, such as calcified, fibrous, atheromatous, thrombotic, metallic and non-diseased regions.

**[0176]** **Figure 25a** shows a schematic representation of an inner border **245** identified by OCT, an outer border **246** identified in the second image by IVUS and a region of interest **247** used for analysis of the OCT and ultrasound signal properties. As shown in **Figure 25b**, the signal properties **248** from the more than one modalities of imaging in the co-registered region of interest are used to generate an assessment of the composition of one or more pixels in the composite image that correspond to the region of interest analyzed. The assessment may be formed by a pattern recognition system **249** trained using methods known in the art. The geometric features **249** identified in the co-registered images are also optionally included in the composite image. The process of assessing the composition of a region of interest can be repeated several times over for different regions of interest to generate a composite image.

**[0177]** In addition, the software and image processing algorithms that enables such analysis of the combined imaging means need not be on the acquisition station. Once the imaging data is acquired, the imaging data can be transferred to allow analysis to occur offline on a separate set of one or more processing units.

**[0178]** The combined IVUS/OCT scanning devices disclosed herein may include a rotary encoder. Further details of optical encoders which may used with the combined IVUS/OCT scanning devices are disclosed in copending application Serial No. 11/..... filed concurrently herewith, entitled

**MEDICAL IMAGING DEVICE WITH ROTARY ENCODER**, which is incorporated herein by reference in its entirety.

**[0179]** Briefly, referring to **Figures 26a to 26e**, the imaging probes may incorporate an encoder which is designed be used with an elongate imaging probe that uses a rotary shaft such as the imaging conduit **34** as part of its scanning mechanism, its use can be generalized for use with any device that makes use of a long, flexible cable used for transmission of torque where non-uniform rotational distortion may occur and an accurate estimation of rotary motion is required. In particular, it is most suited for use with flexible torque transmission systems where the outer diameter of the torque cable is relatively small (e.g. less than 4mm) and long (e.g. longer than 5 cm) such that conventional rotary encoding systems would not provide the desired angular resolution or be adequately compact for the intended use.

**[0180]** **Figure 26a** demonstrates a longitudinal cross-section of the proximal and distal ends of an elongate imaging device **450** with a torque transmission shaft **451**, mechanically coupled to a torque source **452**. The torque source **452** can be a motor, a handle that is manually turned by the operator or any other such device. The

torque transmission shaft **452** transmits torque to the functional end **454** of the device, which can be an energy delivery device, a needle, an atherectomy head or any of several other implements. In **Figure 26c**, the wall of an external sheath **453** is shown to surround the transmission shaft and is shown to enclose the functional end of the device although embodiments where the external sheath is open or has openings near the functional end are possible. An optical fiber **455** is shown to be included as part of the external sheath **453** for the purposes of enabling either the emitting light, detecting light or both to travel to or from the encoding interface **104** that is remote to the proximal end of the transmission sheath. In **Figure 26a** the cylindrical encoding interface body **180** in this case is attached to a rotating portion of the device while the fiber is relatively stationary. The optical fiber **455** may be included as part of the extrusion of the external sheath **453**, as shown, or may be added to the inner or outer surface of the sheath and anchored to the sheath **453** by methods well known in the art, such as bonding or surrounding the fiber and sheath with an additional layer of heat shrinkable material. The optical fiber **455** is terminated with any necessary distal optics **115**, such as an optical spacer, lens and/or deflecting mechanism **172** (such as a prism or mirror) to direct light towards the encoding interface **104**.

**[0181]** The encoding interface **104** in **Figure 26b** is similar to that on the cylindrical encoding interface body in the above mentioned copending application. As the encoding optical circuit used in the embodiments of figures **14a** and **14b** are not mounted onto or directly coupled with the torque transmission shaft, there is no need for an optical rotary joint along the optical encoding circuit.

**[0182]** **Figure 26c** shows a cross-sectional image of a representative cross-section through the device **450** in **Figure 26b** through line **14c-14c**. One or more fiber optics **455** for the encoding system may be incorporated with the external sheath **453**.

**[0183]** Thus the rotary encoder embodiments disclosed in copending application Serial No. 11/..... filed concurrently herewith, entitled **MEDICAL IMAGING DEVICE WITH ROTARY ENCODER**, mentioned above can be incorporated into an imaging probe 12 by substituting the functional end of any of the embodiments in **Figures 26a** to **26d** for an imaging assembly **30** and substituting the torque transmission shaft **451** for an imaging conduit **34** suitable for carrying either electrical or optical signals.

**[0184]** As used herein, the terms "comprises", "comprising", "includes" and "including" are to be construed as being inclusive and open ended, and not exclusive. Specifically, when used in this specification including claims, the terms "comprises", "comprising", "includes" and "including" and variations thereof mean the specified features, steps or components are included. These terms are not to be interpreted to exclude the presence of other features, steps or components.

**[0185]** The foregoing description of the preferred em-

bodiments of the invention has been presented to illustrate the principles of the invention and not to limit the invention to the particular embodiment.

## Claims

1. An imaging probe for insertion into bodily lumens and cavities for imaging an interior of said bodily lumens and cavities, comprising:

a) an elongate hollow shaft (48) having a longitudinal axis having distal and proximal end sections and an elongate midsection, an imaging conduit (34), an imaging assembly (30) being located in said elongate hollow shaft (48) remote from said proximal end section (38), said imaging assembly being connected to a first end of said imaging conduit (34), said imaging conduit (34) extending through the elongate hollow shaft (48) and being connectable through a connector (36) or adaptor (14) at a second end thereof to an image processing and display system (16) through the proximal end section (38), said imaging conduit (34) comprising a fiber optic (40; 403) having a distal end and said imaging assembly (30) including an optical emitter / collector including light directing and receiving means associated with said distal end of a fiber optic (40) for directing light imaging energy out of a distal end of said fiber optic (40) and receiving reflected light imaging energy signals and directing said received reflected light imaging energy signals back to said image processing and display system (16), said light directing and receiving means being one or more mirrors (404) or one or more prisms (397) for deflecting the path of light imaging energy, said imaging assembly (30) including an ultrasound transducer comprising a substrate (402), said transducer having a back surface and comprising a backing on said back surface adapted to attenuate signals generated from the back surface on the substrate (402), and said ultrasound transducer being adapted to emit and receive ultrasound imaging energy and said imaging conduit (34) including an electrical conductor (400) for electrically coupling the ultrasound transducer to an ultrasound signal generator and said image processing and display system (16); wherein said light directing and receiving means and at least part of said distal end of said fiber optic (403) are in, or are within, or are located in/within, or are contained in/within, or reside in/within said backing; and are encompassed, or surrounded, or enclosed by said backing to an extent permitting, or allowing, or providing for the distal end of said fiber optic to enter said backing; and are not located

- in said ultrasound transducer substrate (88, 402) and wherein both the acoustic and optical signals are adapted to be emitted at the same angle relative to the longitudinal axis of the probe and wherein the optical signals are adapted to be emitted through a single optically transmissive channel (407) in said transducer and from the face opposite the backing; and
- b) said imaging assembly (30) including a scanning mechanism configured to deliver said light from the optical emitter / collector and ultrasound from said ultrasound transducer (88) along a path out of said elongate hollow shaft (48), the ultrasound transducer (88) and the optical emitter / collector being positioned and oriented relative to each other to enable accurate co-registering of ultrasound images and optical images during scanning a region of interest;
- c) a drive mechanism for imparting motion to said imaging conduit (34) and said imaging assembly (30); and
- d) a controller being connected to the drive mechanism, and connectable to said image processing and display system (16), and being configured to process the images obtained from ultrasound imaging and optical imaging during scanning of a region of interest and to co-register the ultrasound images and optical images.
2. The imaging probe according to claim 1 wherein said drive mechanism is a rotary drive mechanism for imparting rotational motion around said longitudinal axis.
  3. The imaging probe according to claim 2 wherein said rotary motion occurs at an angular velocity and said angular velocity is adjusted by an adjustment means for said rotational drive mechanism.
  4. The imaging probe according to claim 1 wherein said motion is a reciprocating translational movement of the imaging assembly (30) in the hollow elongate shaft (48).
  5. The imaging probe according to claim 1 wherein said optical emitter / collector and said ultrasound transducer are positioned and oriented relative to each other so that the ultrasound imaging energy and the light imaging energy are received simultaneously from substantially the same direction along substantially the same path of propagation.
  6. The imaging probe according to claim 1 wherein said optical emitter / collector and said ultrasound transducer are positioned and oriented relative to each other to enable imaging of the region of interest with a time delay between obtaining the ultrasound image signals and the optical image signals where the path scanned by a first beam of imaging energy is followed in a predominant manner by a second beam of imaging energy selected, where the first imaging beam is selected from the set of i) a beam of light imaging energy and ii) a beam of ultrasound imaging energy and where the second beam of imaging energy is the imaging beam that was not selected from said set to be the first imaging beam, where said predominant manner is such that during a time interval starting at an initial time point the axis of the second imaging beam does not displace itself from further from the initial axis of the first imaging beam at the said initial time point.
  7. The imaging probe according to claim 1 wherein said optical emitter / collector includes focusing and collection optics for focusing light imaging energy emitted from said distal end of end of said fiber optic (40) to said region of interest being imaged and for collecting light imaging energy reflected therefrom.
  8. The imaging probe according to claim 1 wherein said optical emitter / collector is located in said optically transmissive channel (407) to emit light imaging energy through said optically transmissive channel (407) so that the emitted ultrasound signals and the emitted light imaging energy scan said region of interest at the same time.
  9. The imaging probe according to claim 1 wherein said optical emitter / collector (403) is a first optical emitter / collector (403), including at least a second optical emitter / collector (403) configured to emit light imaging energy at a pre-selected angle with respect to the first optical emitter / collector (403), so that light imaging energy emitted by the first optical emitter / collector (403) and light imaging energy emitted by the second optical emitter / collector (403) scan said region of interest at different times.
  10. The imaging probe according to any one of claims 1 to 9 wherein the light imaging energy is usable to enable imaging by optical coherence tomography.
  11. The imaging probe according to any one of claims 1 to 9 wherein the light imaging energy is visible light.
  12. The imaging probe according to any one of claims 1 to 9 wherein the light imaging energy is infrared light.
  13. The imaging probe according to any one of claims 1 to 12 including said image processing and display system (16) and a rotary encoder mechanism coupled to said imaging assembly (30), and wherein rotational motion of the imaging assembly (30) is detected by said rotary encoder, and wherein said image processing and display system (16) is configured to use said detected rotational motion to infer

an imaging angle.

### Patentansprüche

1. Bildgebungssonde zur Einführung in Körperlumina und -höhlen zur Bildgebung eines Inneren der Körperlumina und -höhlen, umfassend:

a) eine längliche Hohlwelle (48) mit einer Längsachse mit distalen und proximalen Endabschnitten und einem länglichen Mittelabschnitt, einem Bildgebungskanal (34), einer Bildgebungsgruppe (30) in der länglichen Hohlwelle (48) fernab dem proximalen Endabschnitt (38), wobei die Bildgebungsgruppe mit einem ersten Ende des Bildgebungskanals (34) verbunden ist, wobei der Bildgebungskanal (34) sich durch die längliche Hohlwelle (48) erstreckt und über ein Verbindungselement (36) oder einen Adapter (14) an einem zweiten Ende davon an ein Bildverarbeitungs- und -anzeigesystem (16) über den proximalen Endabschnitt (38) anschließbar ist, wobei der Bildgebungskanal (34) eine Faseroptik (40, 403) mit einem distalen Ende umfasst, und wobei die Bildgebungsgruppe (30) einen optischen Emitter/Kollektor umfasst, der Lichtleitungs- und -empfangsmittel, die dem distalen Ende einer Faseroptik (40) zugeordnet sind, um Licht-Bildgebungsenergie aus einem distalen Ende der Faseroptik (40) zu leiten und zurückgestrahlte Bildgebungs-Lichtenergiesignale zu empfangen und diese empfangenen Bildgebungs-Lichtenergiesignale zum Bildverarbeitungs- und -anzeigesystem (16) zurückzuleiten, wobei es sich bei den Lichtleitungs- und -empfangsmitteln um einen oder mehrere Spiegel (404) oder ein oder mehrere Prismen (397) zur Ablenkung des Wegs der Licht-Bildgebungsenergie handelt, wobei die Bildgebungsgruppe (30) einen Ultraschall-Wandler umfasst, der ein Substrat (402) umfasst, wobei der Wandler eine Rückseite hat und einen Träger auf der Rückseite umfasst, der zur Dämpfung der von der Rückseite des Substrats (402) erzeugten Signale geeignet ist, und wobei der Ultraschall-Wandler geeignet ist, Ultraschall-Bildgebungsenergie zu emittieren und zu empfangen, und wobei der Bildgebungskanal (34) einen elektrischen Leiter (400) zur elektrischen Kopplung des Ultraschall-Wandler mit einem Ultraschall-Signalgenerator und dem Bildverarbeitungs- und -anzeigesystem (16) umfasst; wobei die Lichtleitungs- und -empfangsmittel und mindestens ein Teil des distalen Endes der Faseroptik (403) im oder innerhalb des Trägers sind, oder im/innerhalb von diesem lokalisiert sind, oder im/innerhalb von diesem enthalten sind, im/in-

nerhalb von diesem verortet sind, und vom Träger in dem Umfang umreifen oder umgeben oder umschlossen sind, der er ermöglicht oder erlaubt, dass das distale Ende der Faseroptik in den Träger eintritt; und nicht im Substrat (88, 402) des Ultraschall-Wandlers lokalisiert sind, und wobei sowohl die akustischen als auch die optischen Signale geeignet sind, relativ zur Längsachse der Sonde im gleichen Winkel emittiert zu werden und wobei die optischen Signal geeignet sind, über einen einzelnen optisch durchlässigen Kanal (407) und von einer dem Träger gegenüberliegenden Oberfläche im Wandler emittiert zu werden; und

b) wobei die Bildgebungsgruppe (30) einen Abtastmechanismus umfasst, der derart konfiguriert ist, dass er Licht vom optischen Emitter/Kollektor und Ultraschall vom Ultraschall-Wandler (88) entlang einem Weg aus der länglichen Hohlwelle (48) führt, wobei der Ultraschall-Wandler (88) und der optische Emitter/Kollektor relativ zueinander derart positioniert und orientiert sind, dass eine genaue gemeinsame Aufzeichnung von Ultraschallbildern und optischen Bildern bei der Abtastung eines interessierenden Bereichs ermöglicht wird;

c) einen Antriebsmechanismus zur Übertragung von Bewegung auf den Bildgebungskanal (34) und die Bildgebungsgruppe (30); und

d) wobei eine Steuerung an den Antriebsmechanismus angeschlossen ist und an das Bildverarbeitungs- und -anzeigesystem (16) anschließbar ist, und derart konfiguriert ist, dass sie die von der Ultraschall- und optischen Bildgebung bei der Abtastung eines interessierenden Bereichs aufgenommen Bilder verarbeitet und die Ultraschall- und optischen Bilder gemeinsam aufzeichnet.

5

10

15

20

25

30

35

40

45

50

55

2. Bildgebungssonde nach Anspruch 1, wobei der Antriebsmechanismus ein Drehantriebsmechanismus zur Übertragung von Rotation um die Längsachse ist.

3. Bildgebungssonde nach Anspruch 2, wobei die Rotation bei einer Winkelgeschwindigkeit erfolgt und die Winkelgeschwindigkeit durch Einstellmittel des Drehantriebsmechanismus eingestellt wird.

4. Bildgebungssonde nach Anspruch 1, wobei die Bewegung eine hin- und hergehende Translation der Bildgebungsgruppe (30) in der länglichen Hohlwelle (48) ist.

5. Bildgebungssonde nach Anspruch 1, wobei der optische Emitter/Kollektor und der Ultraschall-Wandler relativ zueinander derart positioniert und orientiert sind, dass die Ultraschall-Bildgebungsenergie und



die Licht-Bildgebungsenergie gleichzeitig aus der im Wesentlichen gleichen Richtung und entlang dem im Wesentlichen gleichen Ausbreitungsweg empfangen werden.

6. Bildgebungssonde nach Anspruch 1, wobei der optische Emitter/Kollektor und der Ultraschall-Wandler relativ zueinander derart positioniert und orientiert sind, dass die Bildgebung des interessierenden Bereichs mit einer Zeitverzögerung zwischen der Aufnahme der Ultraschall-Bildsignale und der optischen Bildsignale erfolgt, wobei auf den von einem ersten Strahl von Bildgebungsenergie abgetasteten Weg auf eine vorherrschende Weise ein zweiter ausgewählter Strahl von Bildgebungsenergie folgt, wobei der erste Bildgebungsstrahl aus der nachfolgenden Gruppe gewählt ist: i) ein Strahl von Licht-Bildgebungsenergie und ii) ein Strahl von Ultraschall-Bildgebungsenergie, und wobei der zweite Strahl von Bildgebungsenergie der Bildgebungsstrahl ist, der aus der Gruppe als erster Bildgebungsstrahl nicht ausgewählt wurde, wobei die vorherrschende Weise derart ist, dass in einem Zeitabstand, der zu einem anfänglichen Zeitpunkt beginnt, die Achse des zweiten Bildgebungsstrahls sich zur anfänglichen Achse des ersten Bildgebungsstrahls zum anfänglichen Zeitpunkt nicht versetzt.
7. Bildgebungssonde nach Anspruch 1, wobei der optische Emitter/Kollektor Fokussier- und Kollektionsoptiken zum Fokussieren der aus dem distalen Ende der Faseroptik (40) in den interessierenden, abgebildeten Bereich emittierten Licht-Bildgebungsenergie und zur Aufnahme der davon zurückgestrahlten Licht-Bildgebungsenergie umfasst.
8. Bildgebungssonde nach Anspruch 1, wobei der optische Emitter/Kollektor im optisch durchlässigen Kanal (407) angeordnet ist, um Licht-Bildgebungsenergie durch den optisch durchlässigen Kanal (407) zu emittieren, damit die emittierten Ultraschallsignale und die emittierte Licht-Bildgebungsenergie den interessierenden Bereich gleichzeitig abtasten.
9. Bildgebungssonde nach Anspruch 1, wobei der optische Emitter/Kollektor (403) ein erster optischer Emitter/Kollektor (403) ist, umfassend mindestens einen zweiten optischen Emitter/Kollektor (403), der derart konfiguriert ist, dass er Licht-Bildgebungsenergie in einem im Voraus gewählten Winkel relativ zum ersten optischen Emitter/Kollektor (403) emittiert, damit die vom ersten optischen Emitter/Kollektor (403) emittierte Licht-Bildgebungsenergie und die vom zweiten optischen Emitter/Kollektor (403) emittierte Licht-Bildgebungsenergie den interessierenden Bereich zu unterschiedlichen Zeitpunkten abtasten.

5

10

15

20

25

30

35

40

45

50

55

10. Bildgebungssonde nach einem der Ansprüche 1 - 9, wobei die Licht-Bildgebungsenergie benutzt werden kann, um Bildgebung mittels optischer Kohärenztomographie zu ermöglichen.

11. Bildgebungssonde nach einem der Ansprüche 1 - 9, wobei die Licht-Bildgebungsenergie sichtbares Licht ist.

12. Bildgebungssonde nach einem der Ansprüche 1 - 9, wobei die Licht-Bildgebungsenergie Infrarot-Licht ist.

13. Bildgebungssonde nach einem der Ansprüche 1 - 12, umfassend das Bildverarbeitungs- und -anzeigesystem (16) und einen mit der Bildgebungsgruppe (30) gekoppelten Drehgebermechanismus, wobei eine Rotation der Bildgebungsgruppe (30) durch den Drehgeber erkannt wird, und wobei das Bildverarbeitungs- und -anzeigesystem (16) derart konfiguriert ist, dass es von der erkannten Rotation auf einen Bildgebungswinkel schließt.

## Revendications

1. Sonde d'imagerie conçue destinée à être insérée dans des lumières et cavités corporelles pour former une image d'un intérieur desdites lumières et cavités corporelles, comprenant :
- a) une tige creuse allongée (48) ayant un axe longitudinal ayant des sections d'extrémités distale et proximale et une section intermédiaire allongée, un conduit d'imagerie (34), un ensemble d'imagerie (30) étant situé dans ladite tige creuse allongée (48) à distance de ladite section d'extrémité proximale (38), ledit ensemble d'imagerie étant relié à une première extrémité dudit conduit d'imagerie (34), ledit conduit d'imagerie (34) s'étendant à travers la tige creuse allongée (48) et pouvant être relié par le biais d'un connecteur (36) ou adaptateur (14) au niveau d'une seconde extrémité de celui-ci à un système de traitement et d'affichage d'images (16) à travers la section d'extrémité proximale (38), ledit conduit d'imagerie (34) comprenant une fibre optique (40 ; 403) ayant une extrémité distale et ledit ensemble d'imagerie (30) possédant un émetteur/collecteur optique comportant des moyens d'orientation et de réception de lumière associés à ladite extrémité distale d'une fibre optique (40) pour orienter l'énergie d'imagerie lumineuse hors d'une extrémité distale de ladite fibre optique (40) et recevoir des signaux d'énergie d'imagerie lumineuse réfléchis et réorienter lesdits signaux d'énergie d'imagerie lumineuse réfléchis reçus vers ledit système de

traitement et d'affichage d'images (16), lesdits moyens d'orientation et de réception de lumière étant un ou plusieurs miroirs (404) ou un ou plusieurs prismes (397) pour dévier la trajectoire de l'énergie d'imagerie lumineuse, ledit ensemble d'imagerie (30) comportant un transducteur ultrasonore comprenant un substrat (402), ledit transducteur ayant une surface arrière et comprenant un support sur ladite surface arrière adapté pour atténuer les signaux générés depuis la surface arrière sur le substrat (402), et ledit transducteur ultrasonore étant adapté pour émettre et recevoir de l'énergie d'imagerie ultrasonore et ledit conduit d'imagerie (34) comportant un conducteur électrique (400) pour coupler électriquement le transducteur ultrasonore à un générateur de signaux ultrasonores et audit système de traitement et d'affichage d'images (16) ; dans lequel lesdits moyens d'orientation et de réception de lumière et au moins une partie de ladite extrémité distale de ladite fibre optique (403) sont dans, ou sont à l'intérieur de, ou sont situés dans/à l'intérieur de, ou sont contenus dans/à l'intérieur de, ou résident dans ou à l'intérieur dudit support ; et sont englobés, ou entourés par, ou enfermés dans ledit support dans une mesure autorisant, ou permettant ou prévoyant que l'extrémité distale de ladite fibre optique pénètre dans ledit support ; et ne sont pas situés dans ledit substrat de transducteur ultrasonore (88, 402) et dans laquelle les signaux acoustiques et optiques sont tous deux adaptés pour être émis suivant le même angle par rapport à l'axe longitudinal de la sonde et dans laquelle les signaux optiques sont adaptés pour être émis par le biais d'un canal optiquement transmissif unique (407) dans ledit transducteur et depuis la face opposée au support ; et

b) ledit ensemble d'imagerie (30) comportant un mécanisme de balayage configuré pour délivrer ladite lumière depuis l'émetteur/le collecteur optique et l'ultrason depuis ledit transducteur ultrasonore (88) le long d'une trajectoire à l'extérieur de ladite tige creuse allongée (48), le transducteur ultrasonore (88) et l'émetteur/le collecteur optique étant positionnés et orientés l'un par rapport à l'autre pour permettre le co-enregistrement précis d'images ultrasonores et d'images optiques au cours du balayage d'une région d'intérêt ;

c) un mécanisme d'entraînement pour communiquer un mouvement audit conduit d'imagerie (34) et audit ensemble d'imagerie (30) ; et

d) un contrôleur relié au mécanisme d'entraînement, et pouvant être relié audit système de traitement et d'affichage d'images (16), et étant configuré pour traiter les images obtenues de l'imagerie ultrasonore et de l'imagerie optique

au cours du balayage d'une région d'intérêt et pour co-enregistrer les images ultrasonores et les images optiques.

- 5 2. Sonde d'imagerie selon la revendication 1 dans laquelle ledit mécanisme d'entraînement est un mécanisme d'entraînement rotatif permettant de communiquer un mouvement de rotation autour dudit axe longitudinal.
- 10 3. Sonde d'imagerie selon la revendication 2 dans laquelle ledit mouvement de rotation survient à une vitesse angulaire et ladite vitesse angulaire est ajustée à l'aide d'un moyen d'ajustement dudit mécanisme d'entraînement rotationnel.
- 15 4. Sonde d'imagerie selon la revendication 1 dans laquelle ledit mouvement est un mouvement translationnel de va-et-vient de l'ensemble d'imagerie (30) dans la tige creuse allongée (48).
- 20 5. Sonde d'imagerie selon la revendication 1, dans laquelle ledit émetteur/collecteur et ledit transducteur ultrasonore sont positionnés et orientés l'un par rapport à l'autre de telle sorte que l'énergie d'imagerie ultrasonore et l'énergie d'imagerie lumineuse sont reçues simultanément depuis sensiblement la même direction le long de sensiblement la même trajectoire de propagation.
- 25 6. Sonde d'imagerie selon la revendication 1, dans laquelle ledit émetteur/collecteur optique et ledit transducteur ultrasonore sont positionnés et orientés l'un par rapport à l'autre pour permettre l'imagerie de la région d'intérêt avec un temps de retard entre l'obtention des signaux d'images ultrasonores et des signaux d'images optiques où la trajectoire balayée par un premier faisceau d'énergie d'imagerie est suivie de manière prédominante par un second faisceau d'énergie d'imagerie sélectionné, où le premier faisceau d'imagerie est sélectionné dans l'ensemble comprenant i) un faisceau d'énergie d'imagerie lumineuse et ii) un faisceau d'énergie d'imagerie ultrasonore et où le second faisceau d'imagerie est le faisceau d'imagerie qui n'a pas été sélectionné dans ledit ensemble pour être le premier faisceau d'imagerie, où ladite manière prédominante est telle qu'au cours d'un intervalle de temps partant d'un moment initial, l'axe du second faisceau d'imagerie ne se déplace pas lui-même de plus loin par rapport à l'axe initial du premier faisceau d'imagerie audit moment initial.
- 30 7. Sonde d'imagerie selon la revendication 1, dans laquelle ledit émetteur/collecteur optique comporte une optique de focalisation et de collecte pour focaliser l'énergie d'imagerie lumineuse émise par ladite extrémité distale d'extrémité de ladite fibre optique
- 35
- 40
- 45
- 50
- 55

(40) vers ladite région d'intérêt étant imagée et pour collecter l'énergie d'imagerie lumineuse réfléchie par celle-ci.

8. Sonde d'imagerie selon la revendication 1, dans laquelle ledit émetteur/collecteur optique est situé dans ledit canal optiquement transmissif (407) pour émettre l'énergie d'imagerie lumineuse à travers ledit canal optiquement transmissif (407) de telle sorte que les signaux ultrasonores émis et l'énergie d'imagerie lumineuse émise balayent ladite région d'intérêt en même temps. 5  
10
9. Sonde d'imagerie selon la revendication 1, dans laquelle ledit émetteur/collecteur optique (403) est un premier émetteur/collecteur optique (403), comportant au moins un second émetteur/collecteur optique (403) configuré pour émettre de l'énergie d'imagerie lumineuse selon un angle présélectionné par rapport au premier émetteur/collecteur optique (403), de telle sorte que l'énergie d'imagerie lumineuse émise par le premier émetteur/collecteur optique (403) et l'énergie d'imagerie lumineuse émise par le second émetteur/collecteur optique (403) balayent ladite région d'intérêt à des moments différents. 15  
20  
25
10. Sonde d'imagerie selon l'une quelconque des revendications 1 à 9, dans laquelle l'énergie d'imagerie lumineuse est utilisable pour permettre une imagerie par tomographie par cohérence optique. 30
11. Sonde d'imagerie selon l'une quelconque des revendications 1 à 9, dans laquelle l'énergie d'imagerie lumineuse est de la lumière visible. 35
12. Sonde d'imagerie selon l'une quelconque des revendications 1 à 9, dans laquelle l'énergie d'imagerie lumineuse est de la lumière infrarouge.
13. Sonde d'imagerie selon l'une quelconque des revendications 1 à 12, comportant ledit système de traitement et d'affichage d'images (16) et un mécanisme de codage rotatif couplé audit ensemble d'imagerie (30), et dans laquelle le mouvement rotationnel de l'ensemble d'imagerie (30) est détecté par ledit codeur rotatif, et dans laquelle ledit système de traitement et d'affichage d'images (16) est configuré pour utiliser ledit mouvement de rotation détecté pour inférer un angle d'imagerie. 40  
45  
50

55

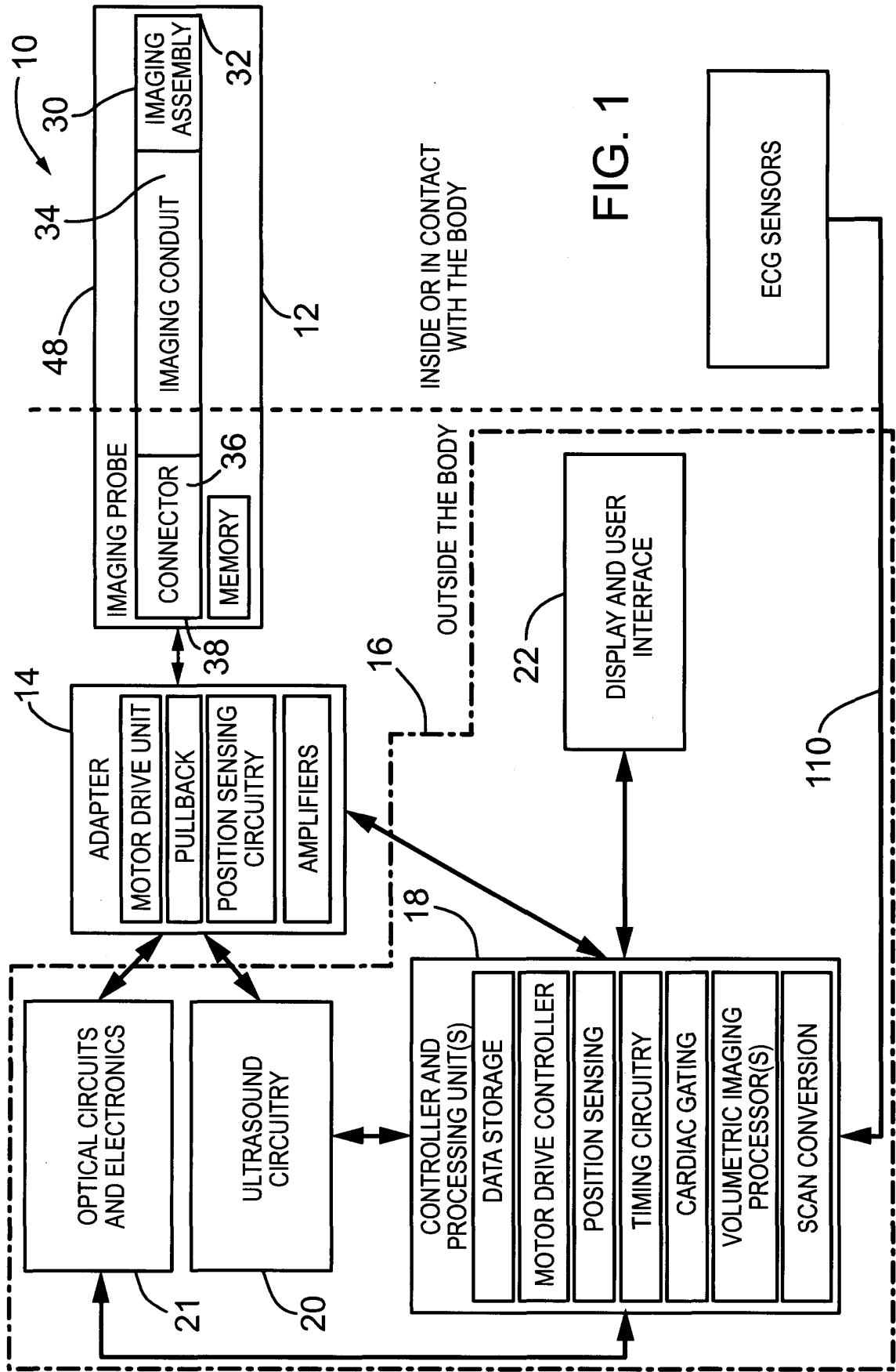


FIG. 1

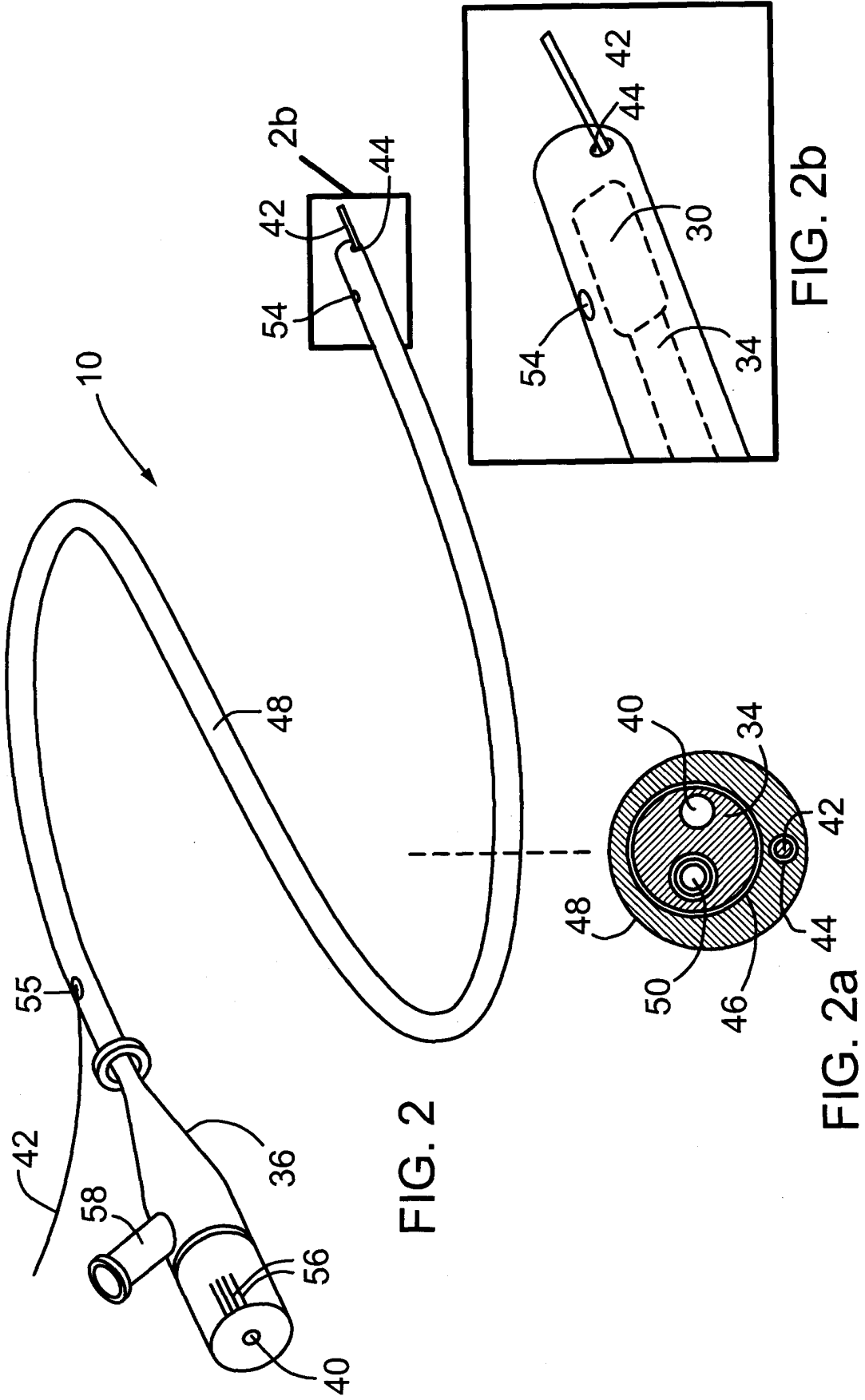
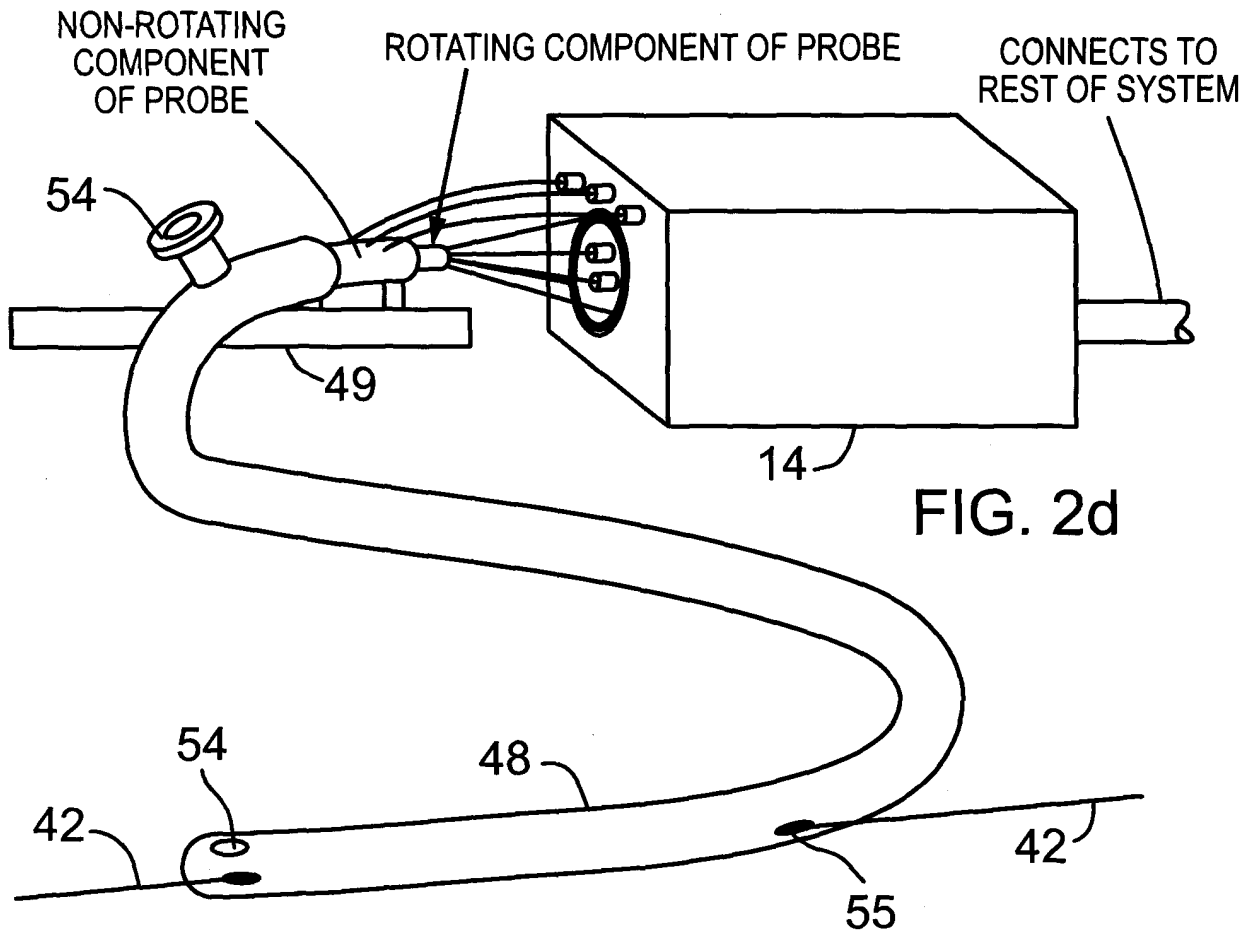
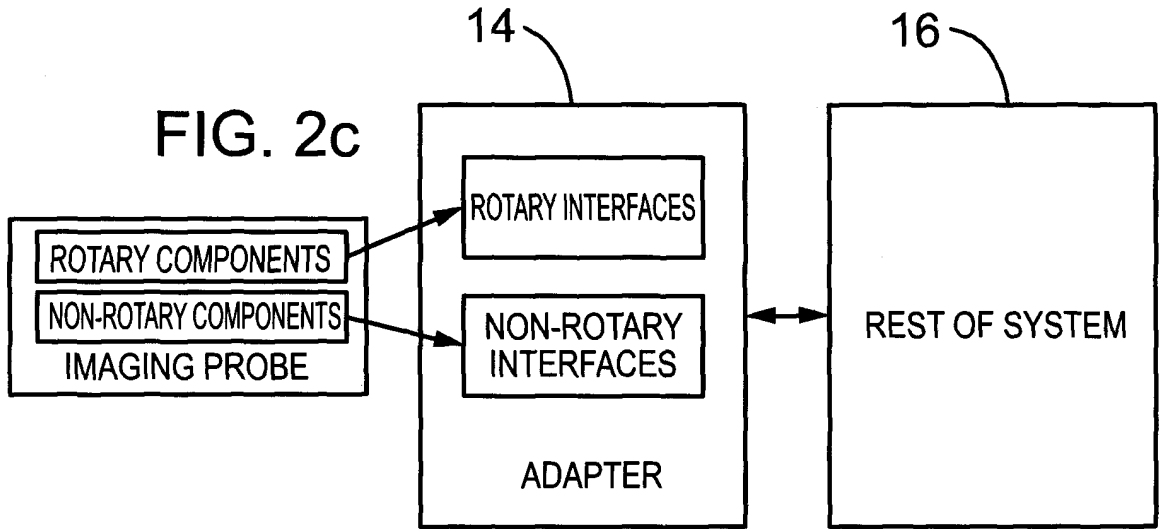


FIG. 2

FIG. 2b

FIG. 2a



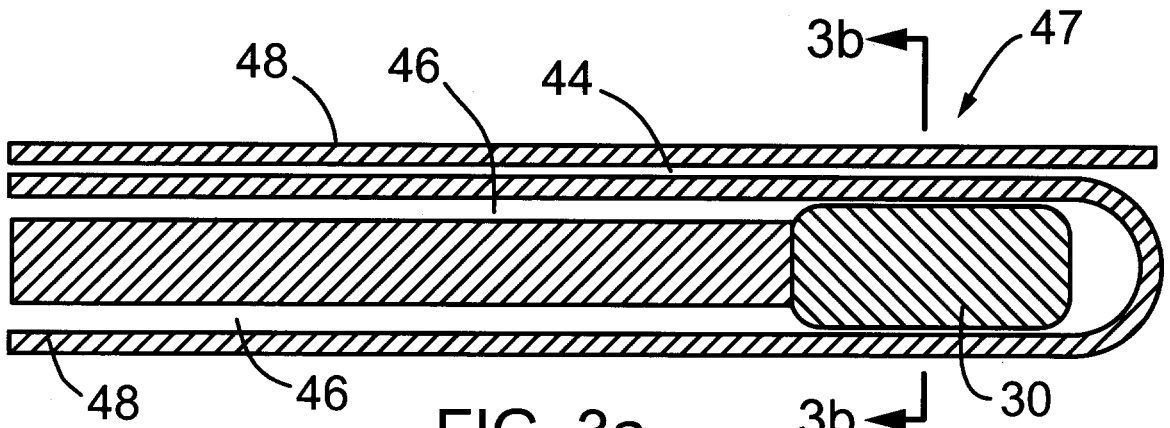


FIG. 3a

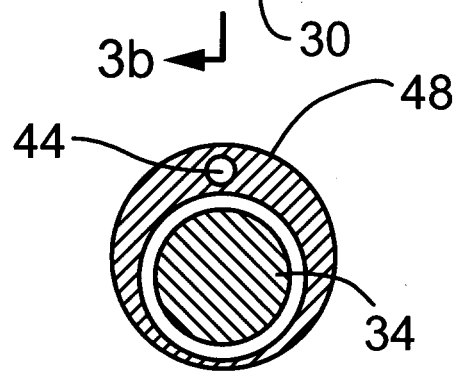


FIG. 3b

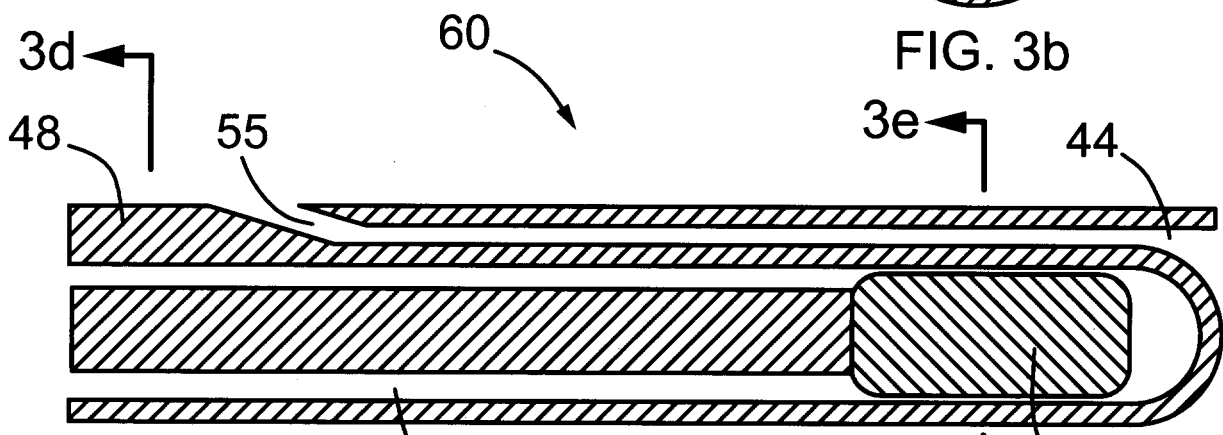


FIG. 3c

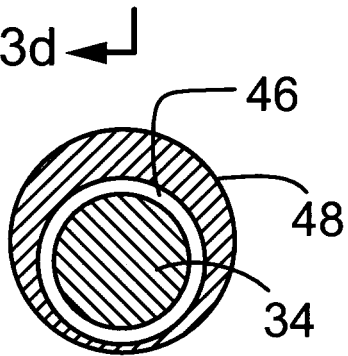


FIG. 3d

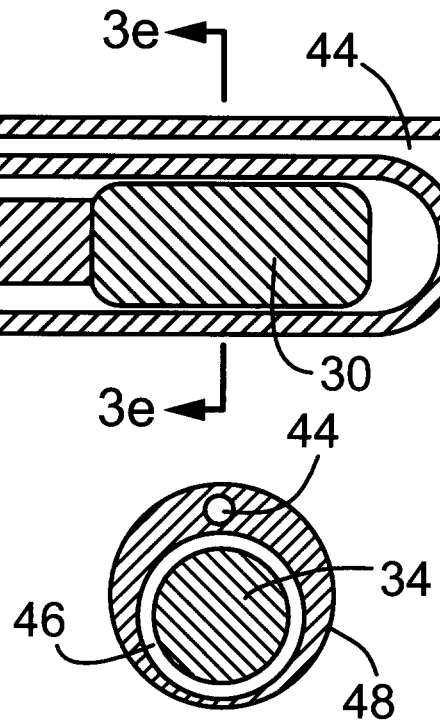
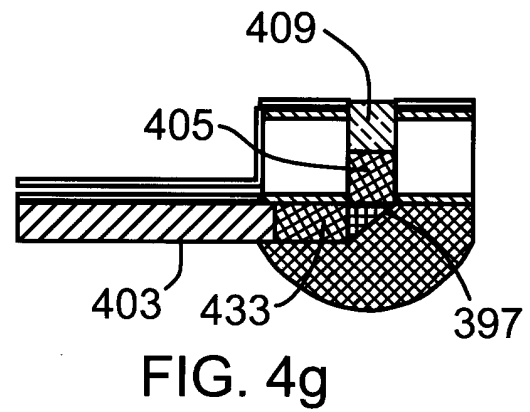
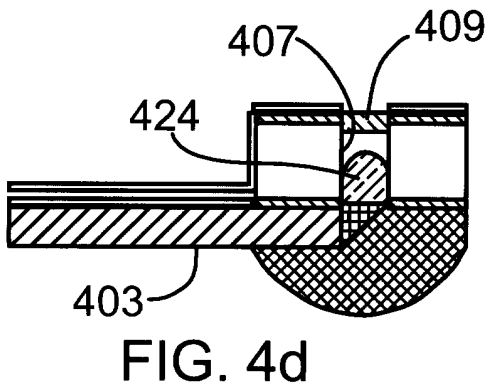
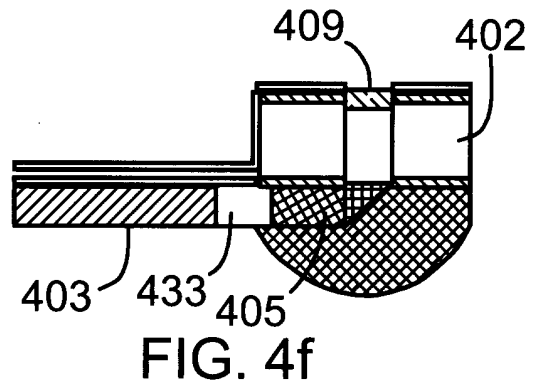
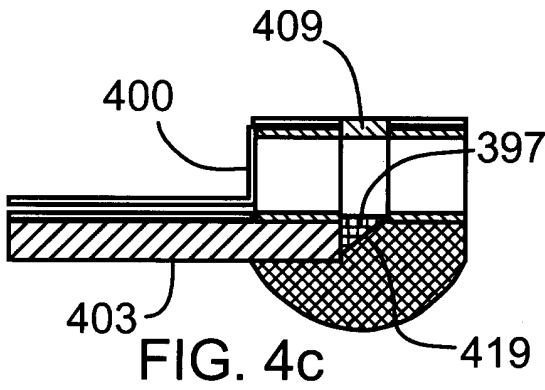
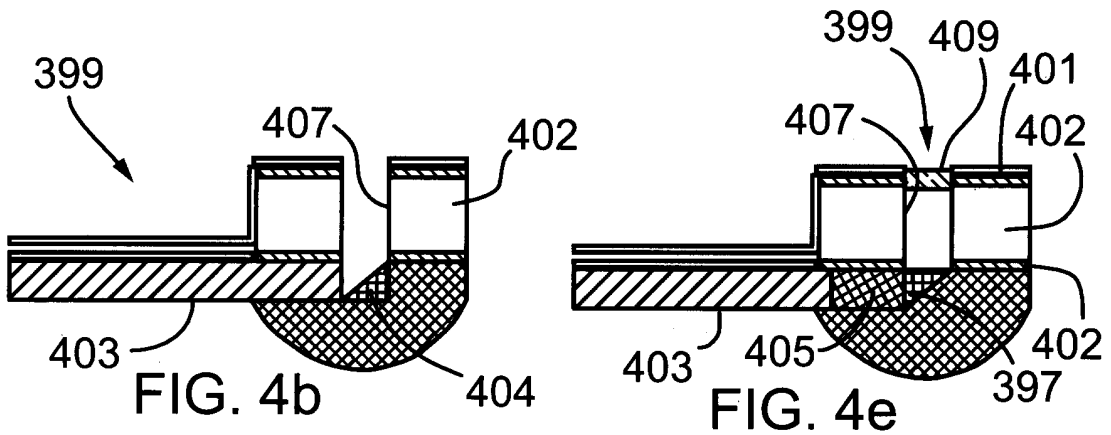
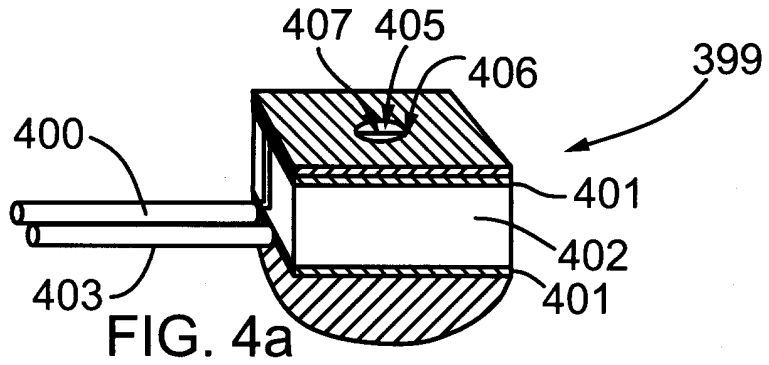


FIG. 3e





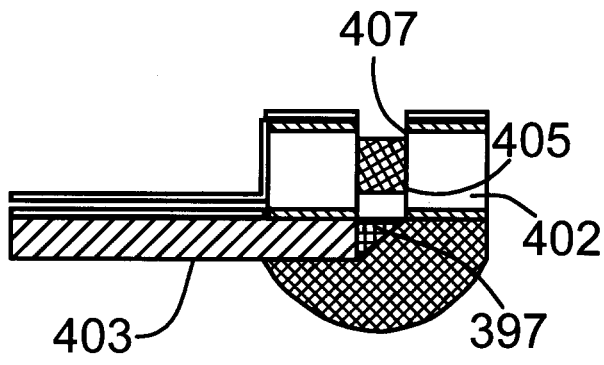


FIG. 4h

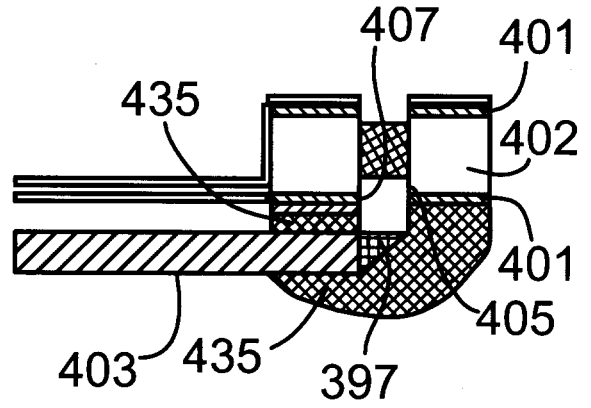


FIG. 4j

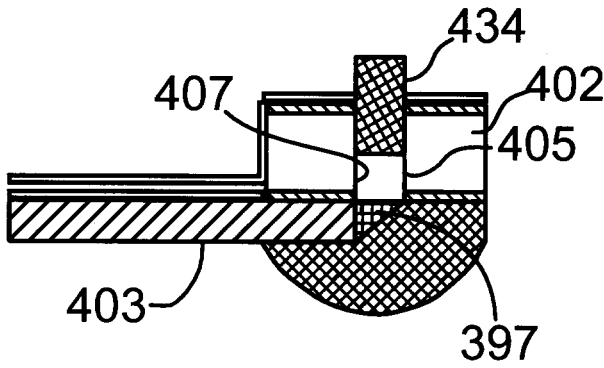


FIG. 4i

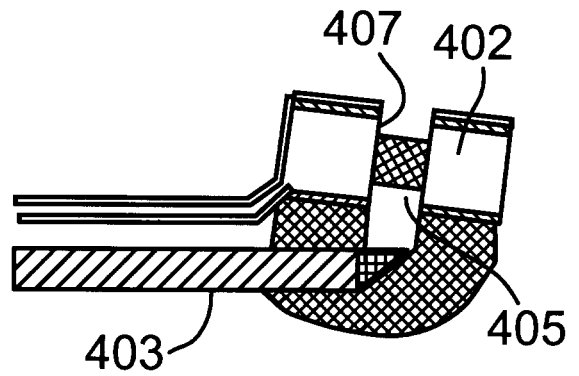


FIG. 4k

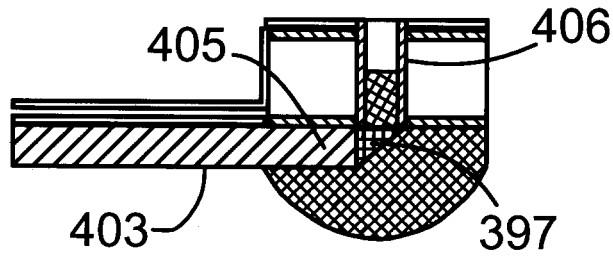


FIG. 4l

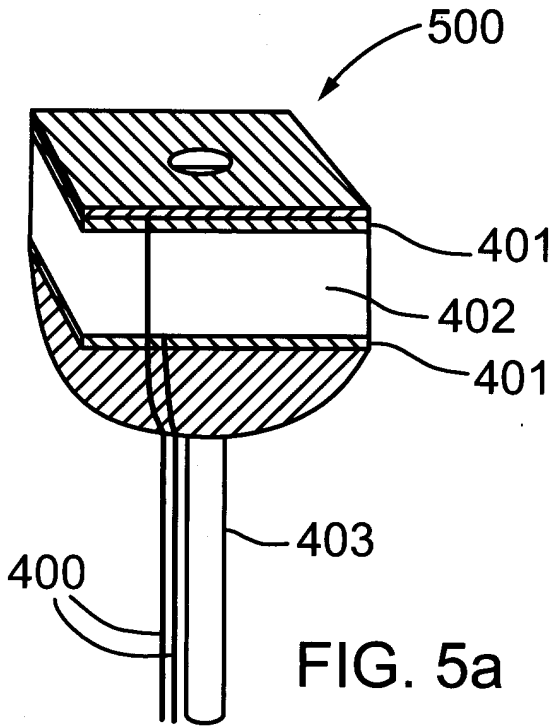


FIG. 5a

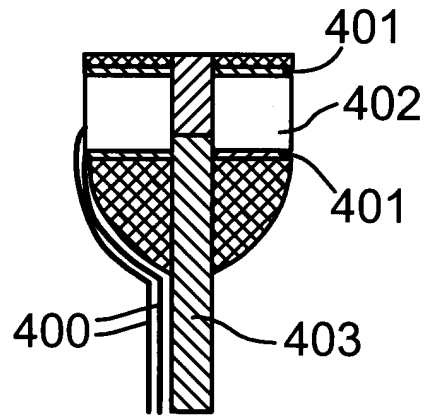


FIG. 5d

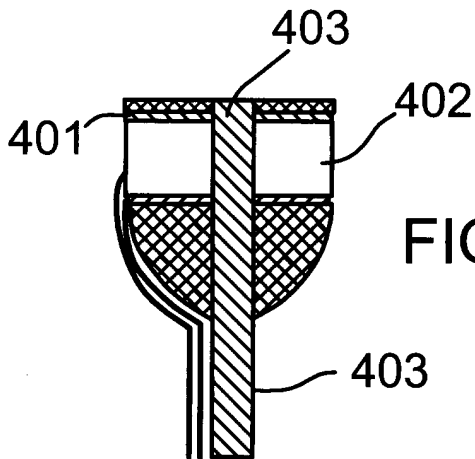


FIG. 5b

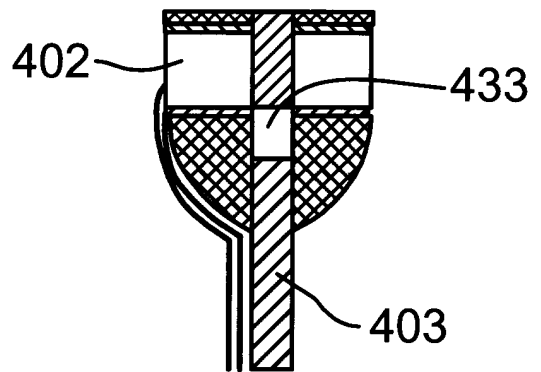


FIG. 5e

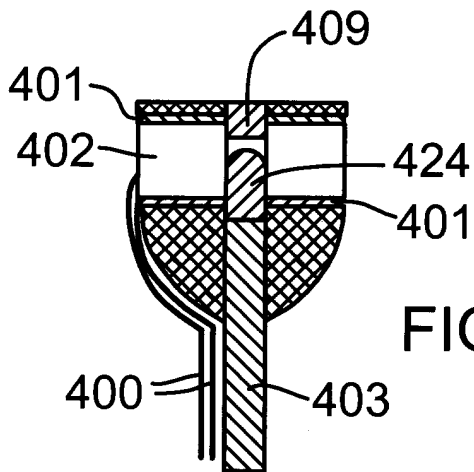


FIG. 5c

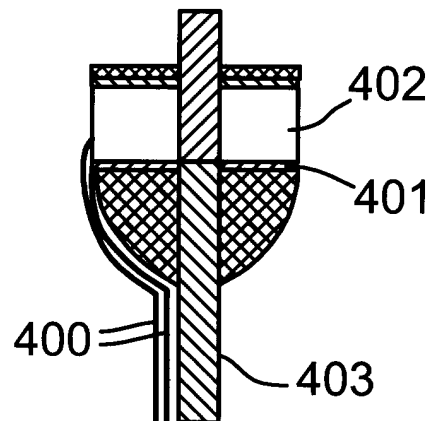


FIG. 5f

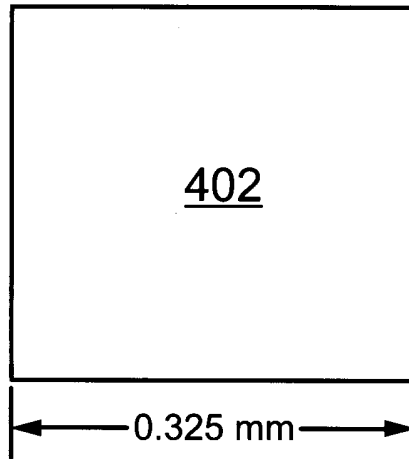


FIG. 6a

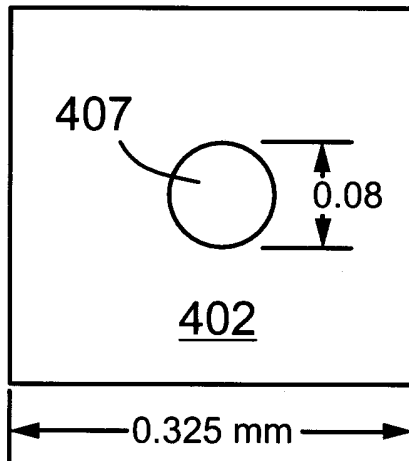


FIG. 6b

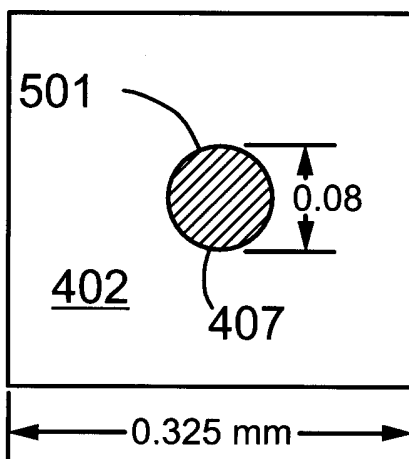
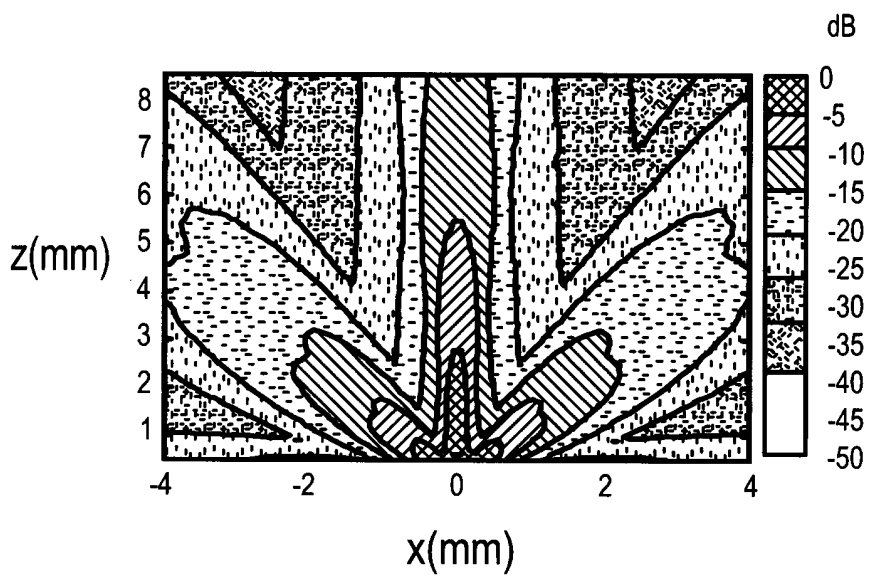
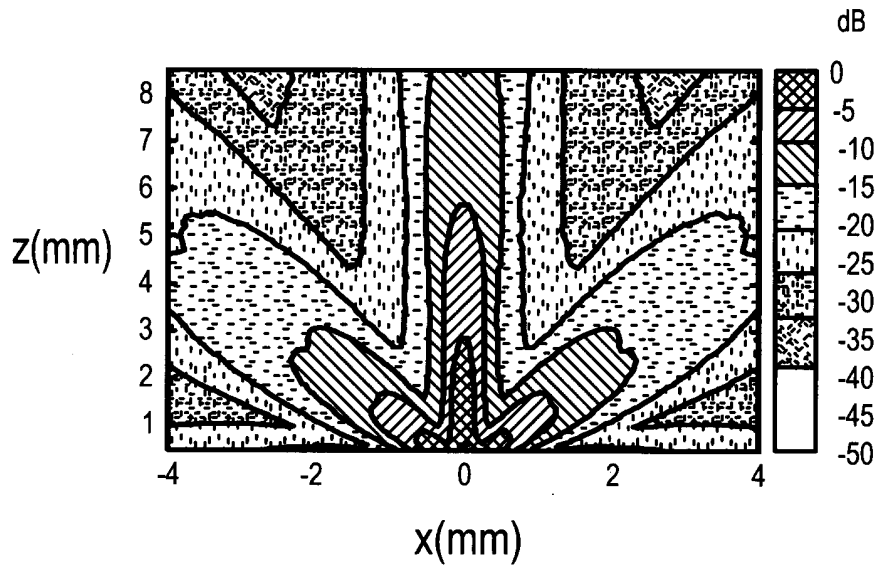
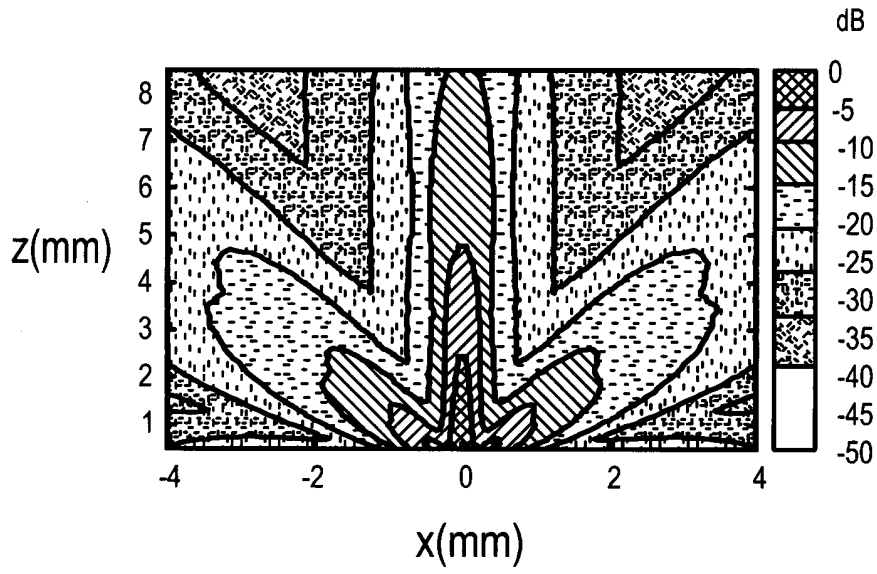


FIG. 6c



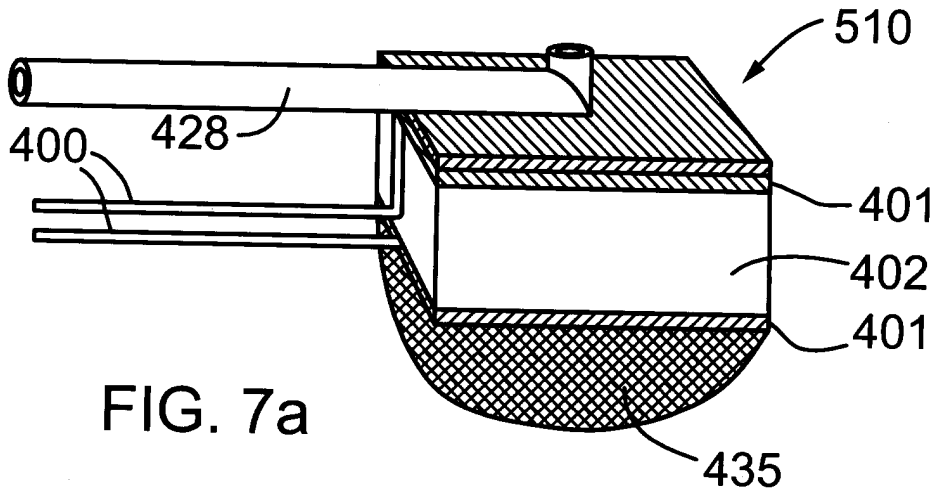


FIG. 7a

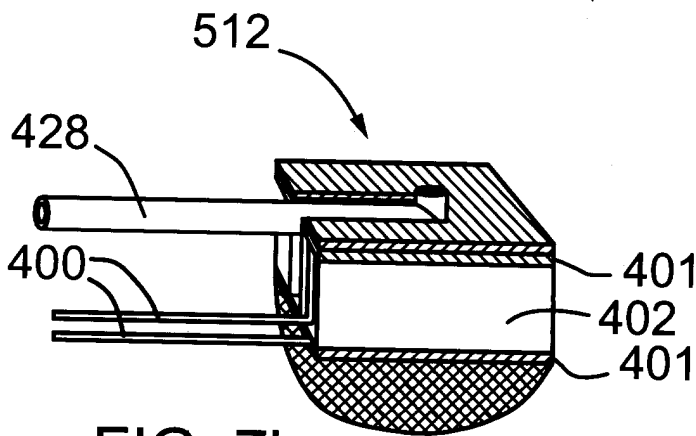


FIG. 7b

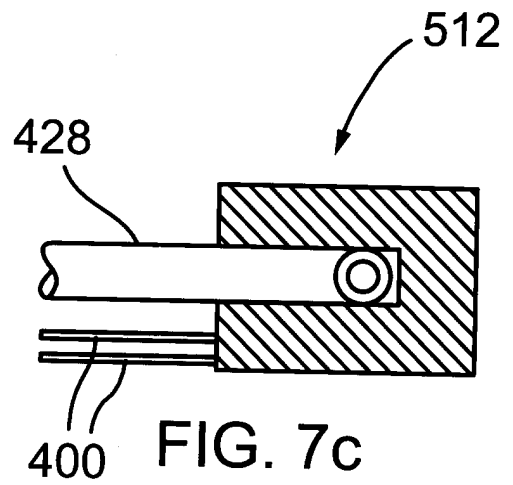


FIG. 7c

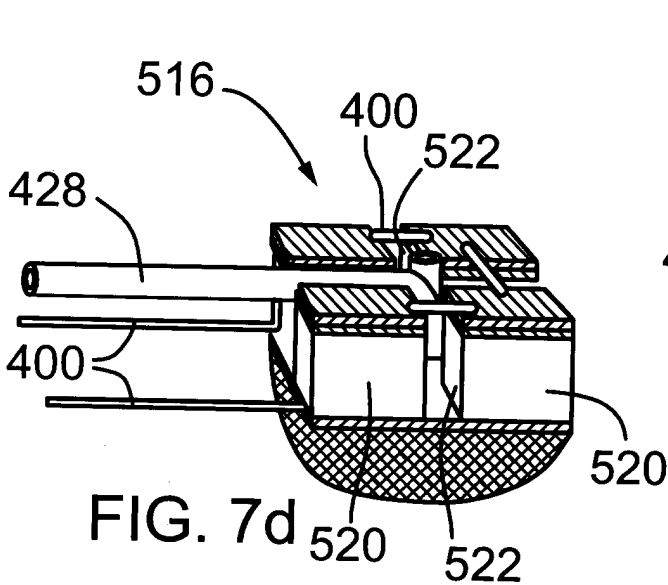


FIG. 7d

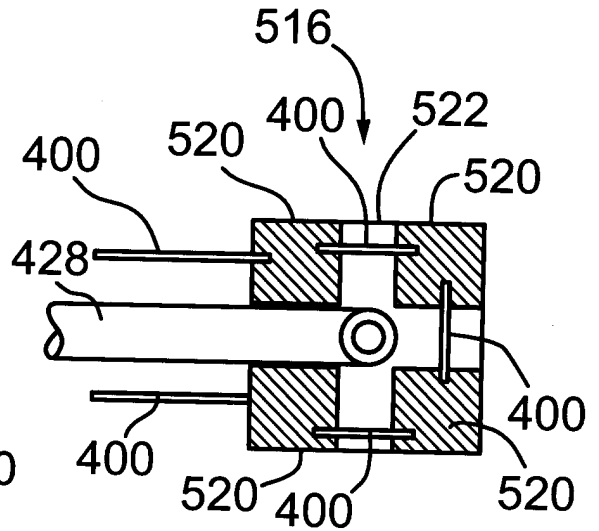
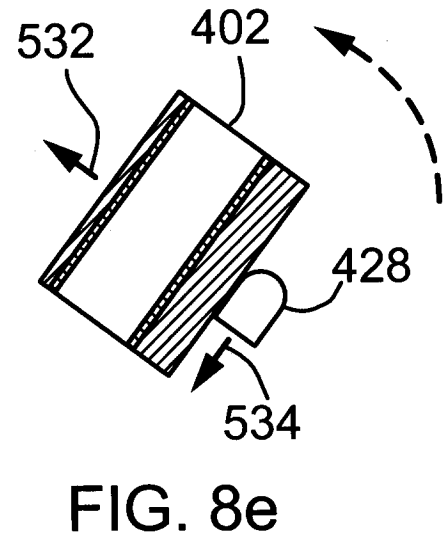
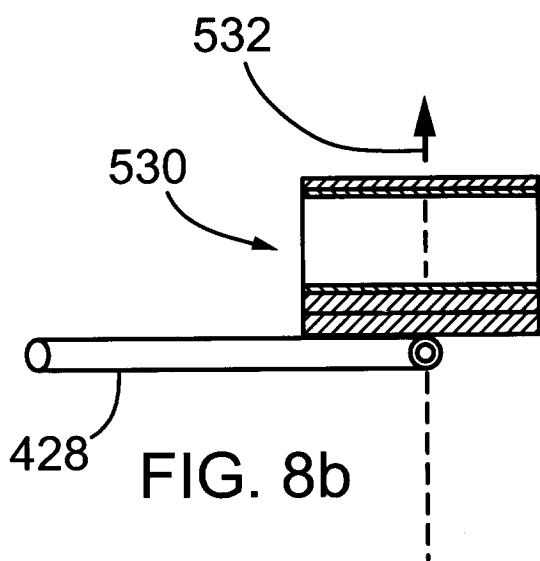
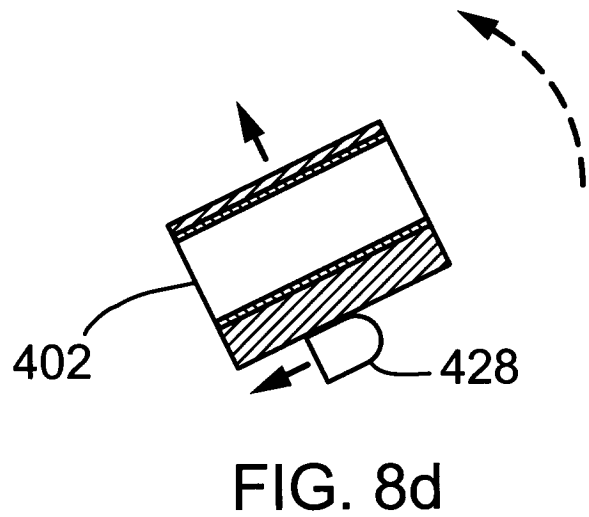
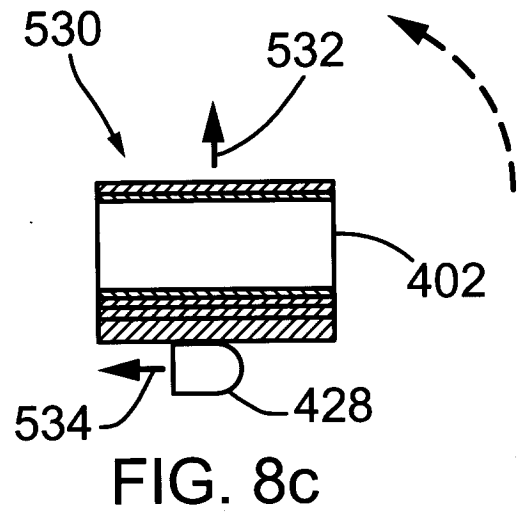
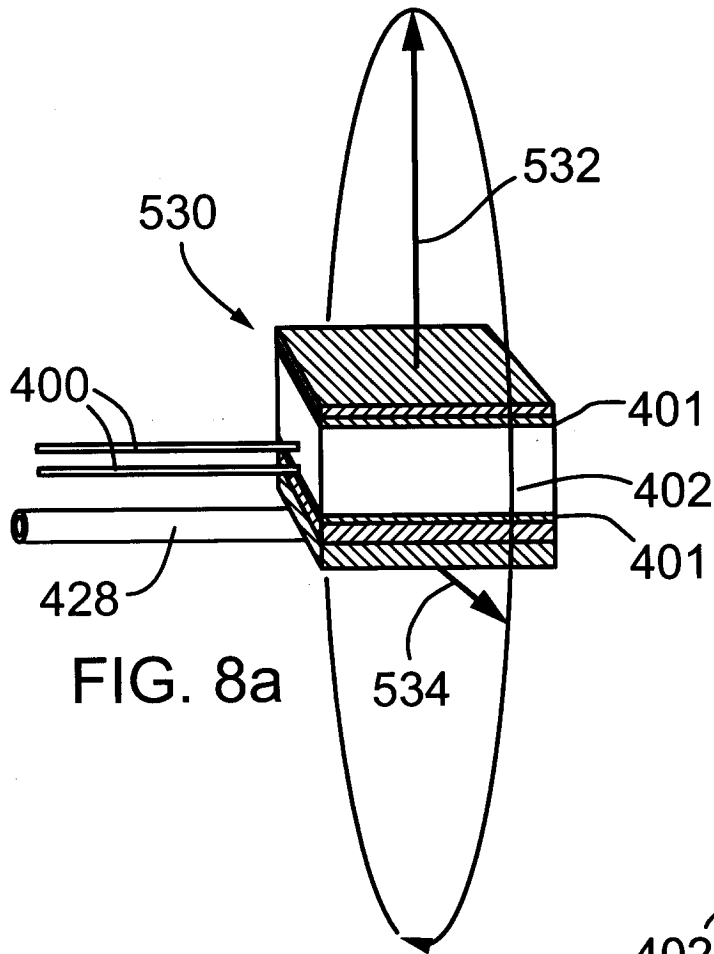


FIG. 7e



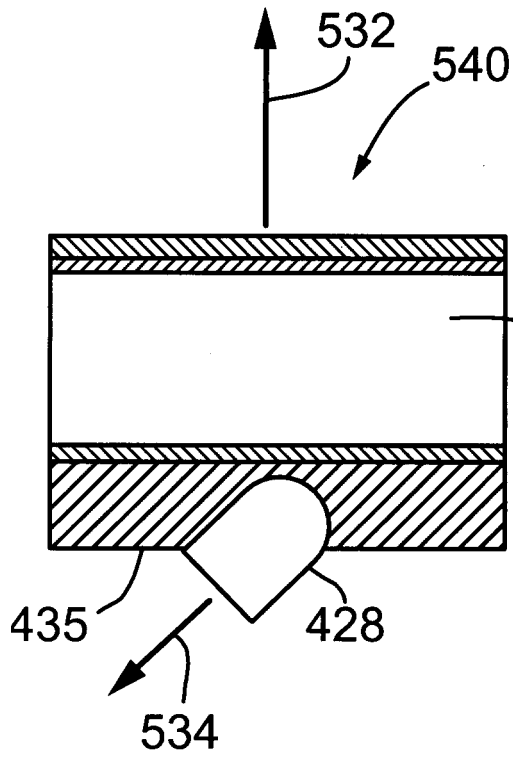


FIG. 9a

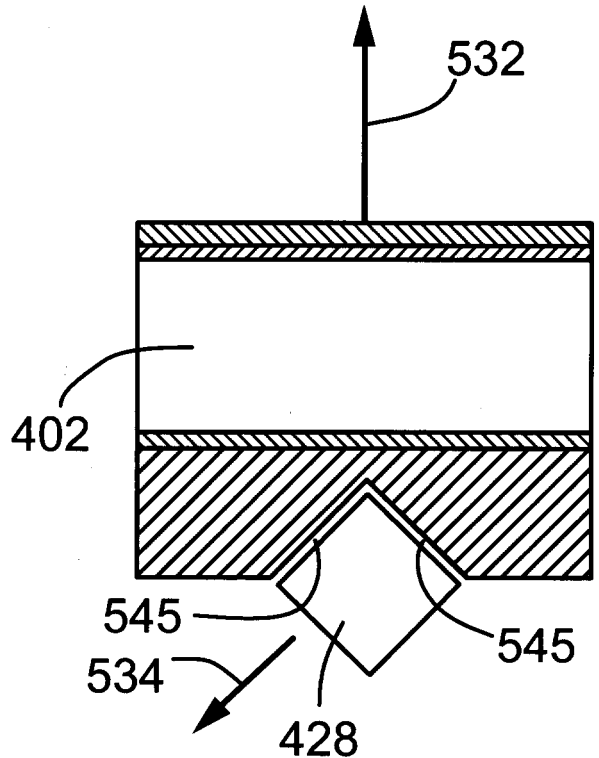


FIG. 9c

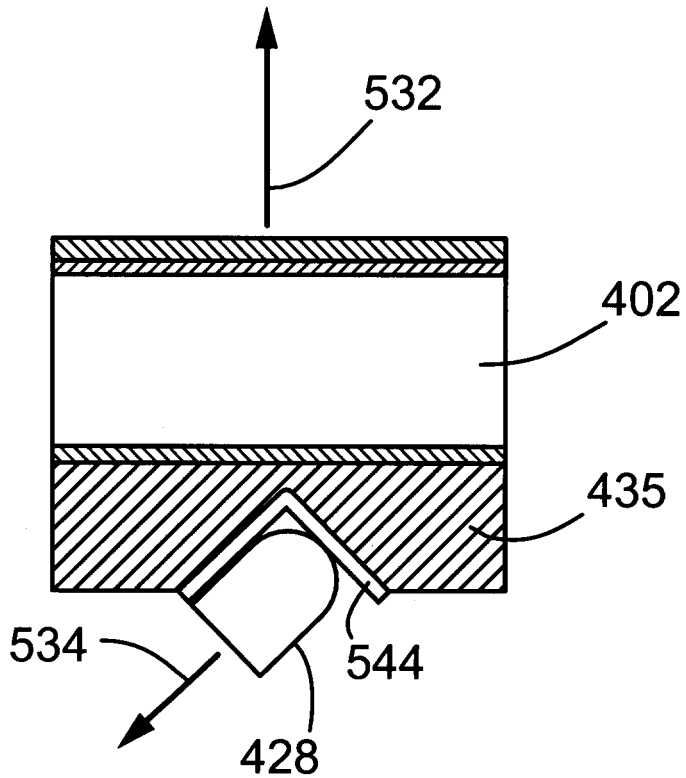
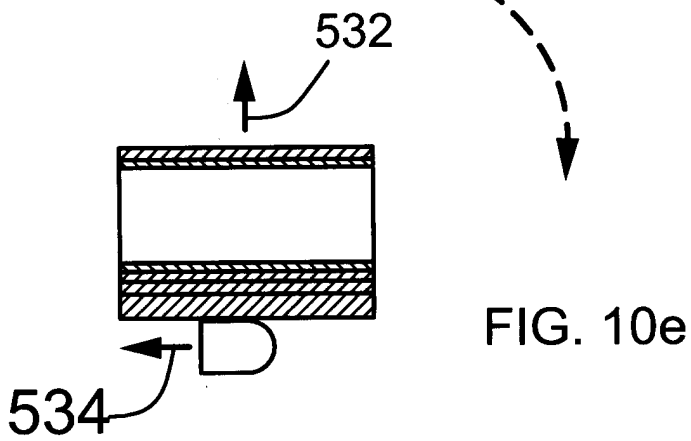
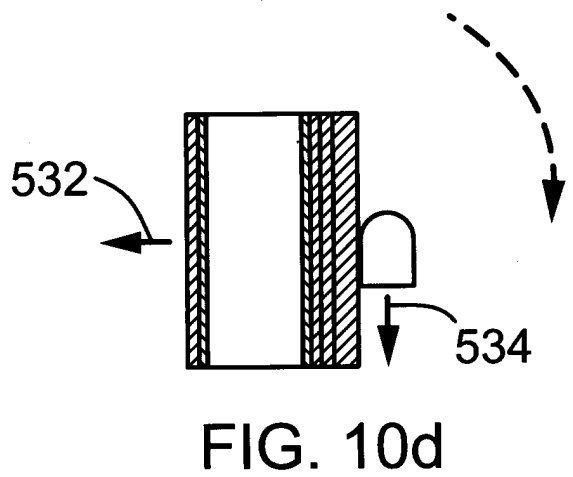
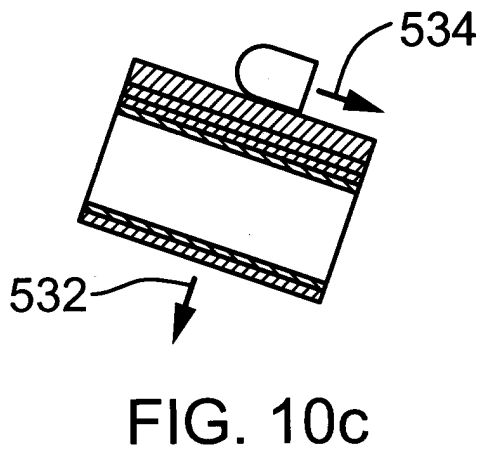
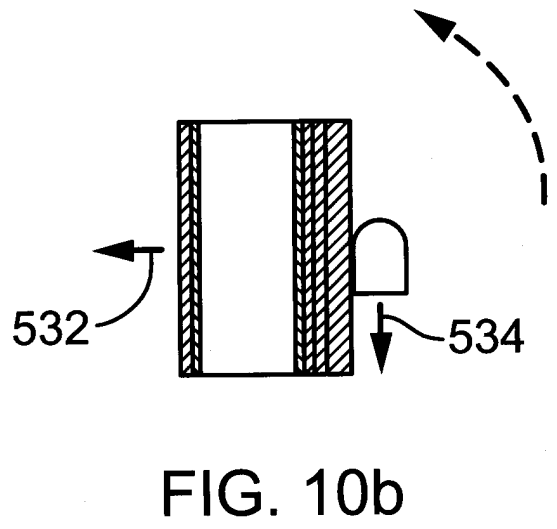
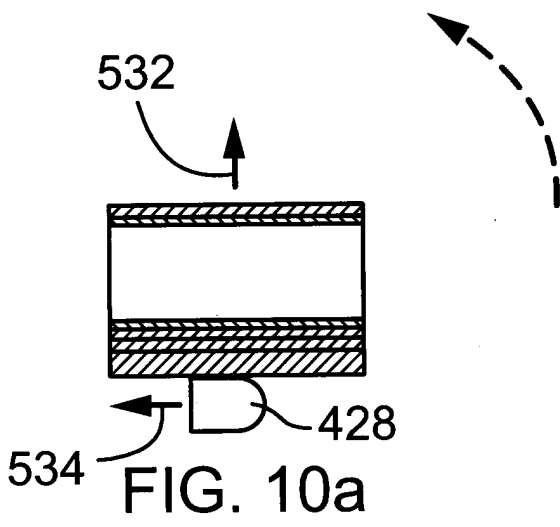
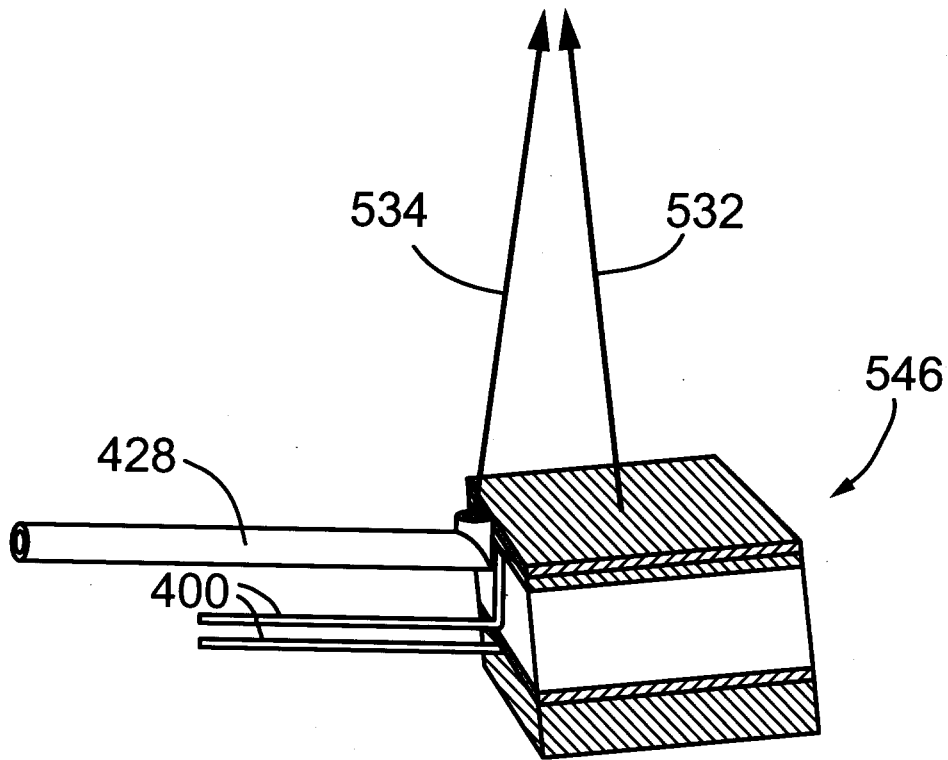
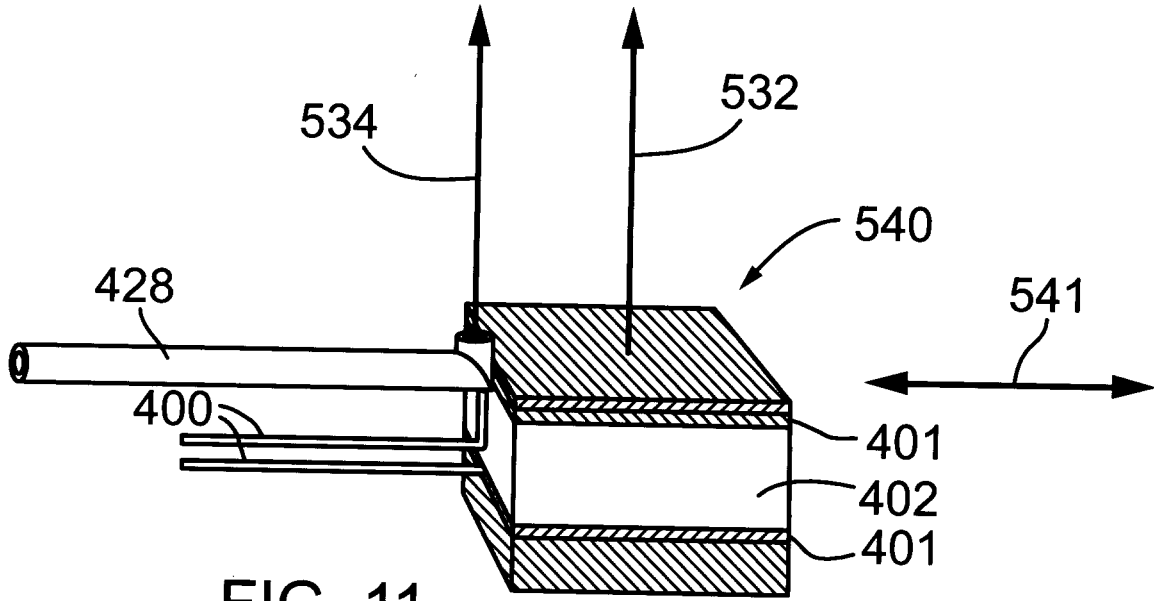


FIG. 9b







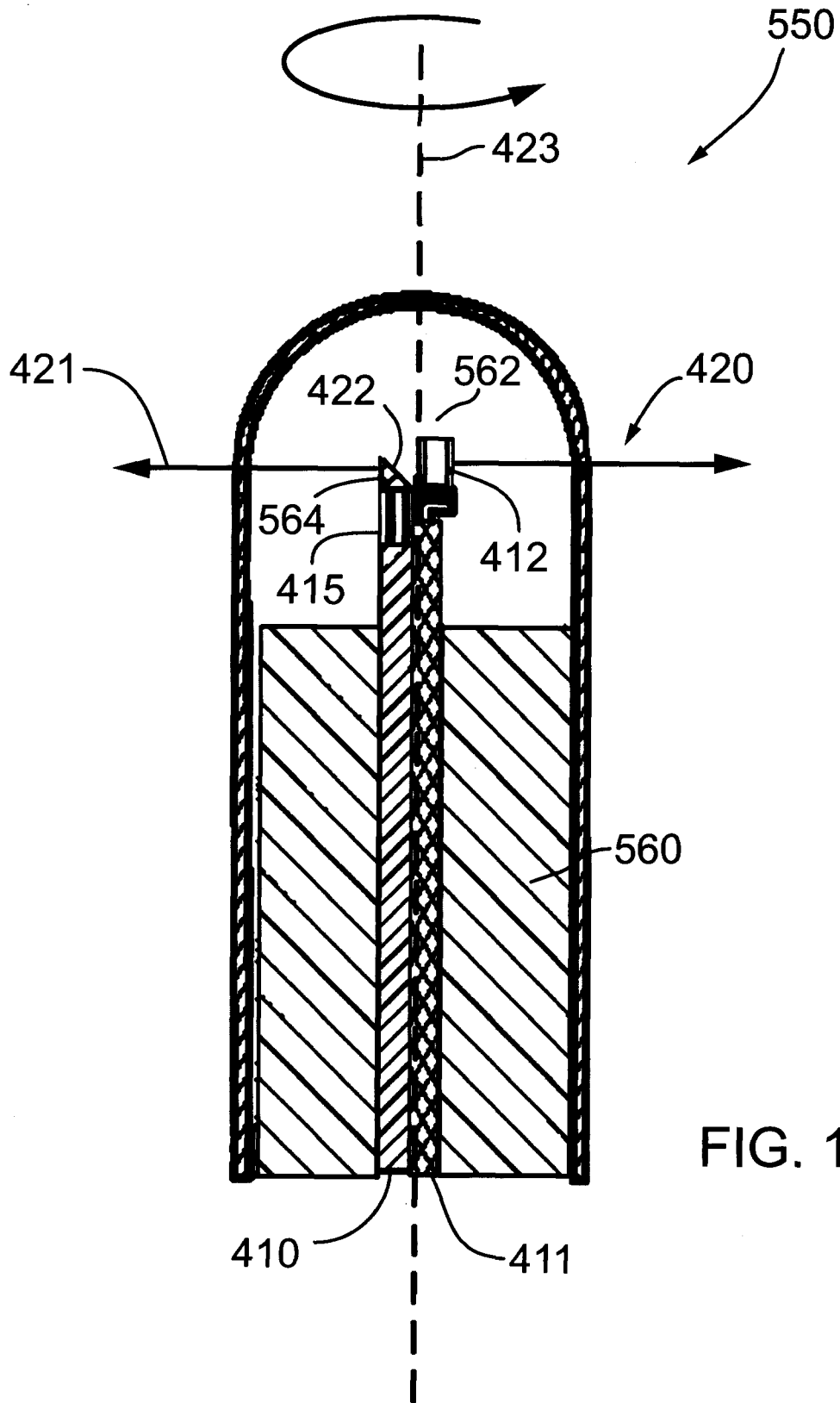


FIG. 13

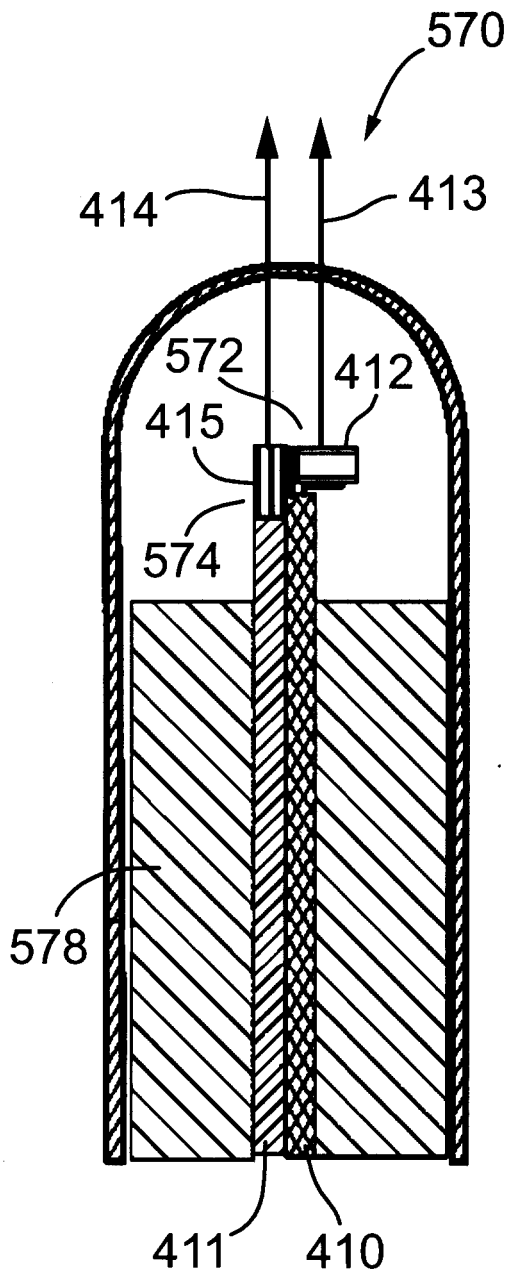


FIG. 14a

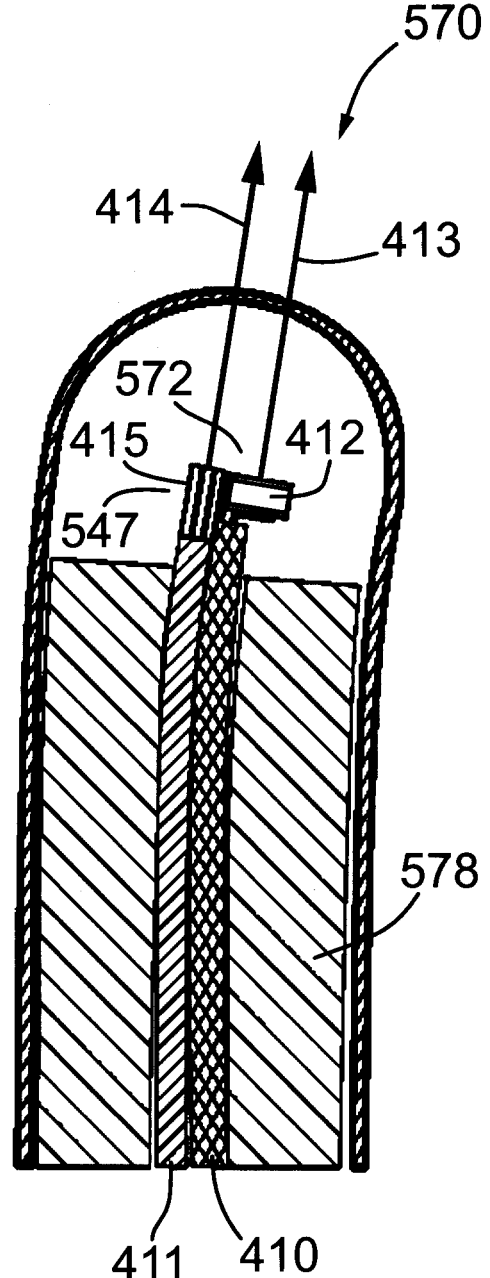


FIG. 14b

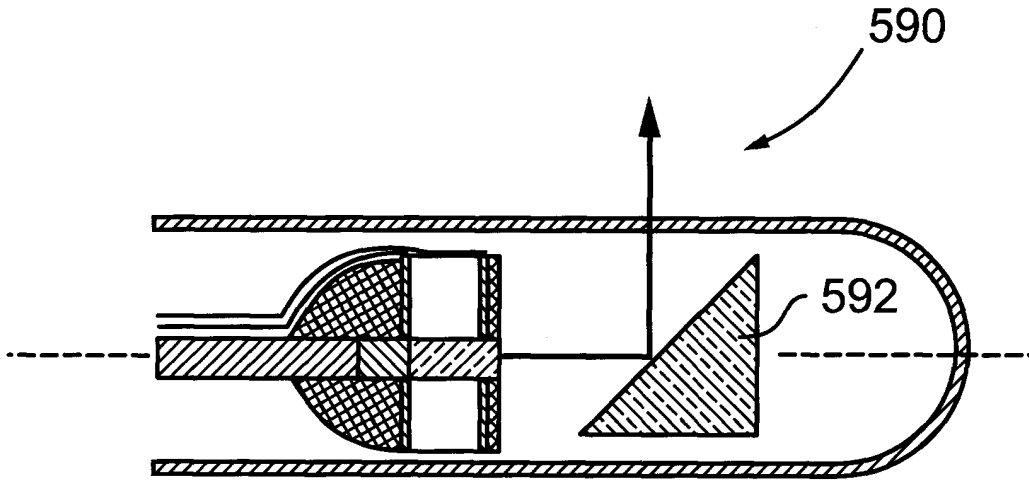


FIG. 15a

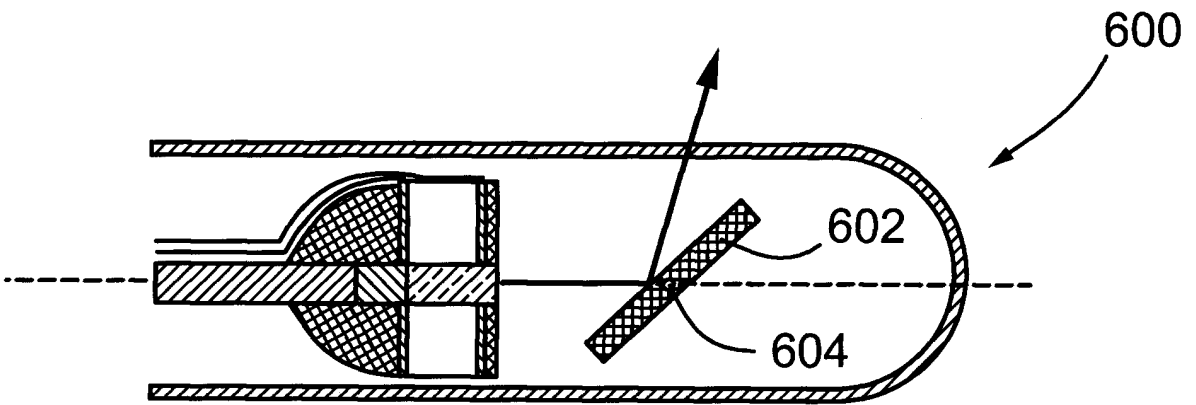


FIG. 15b

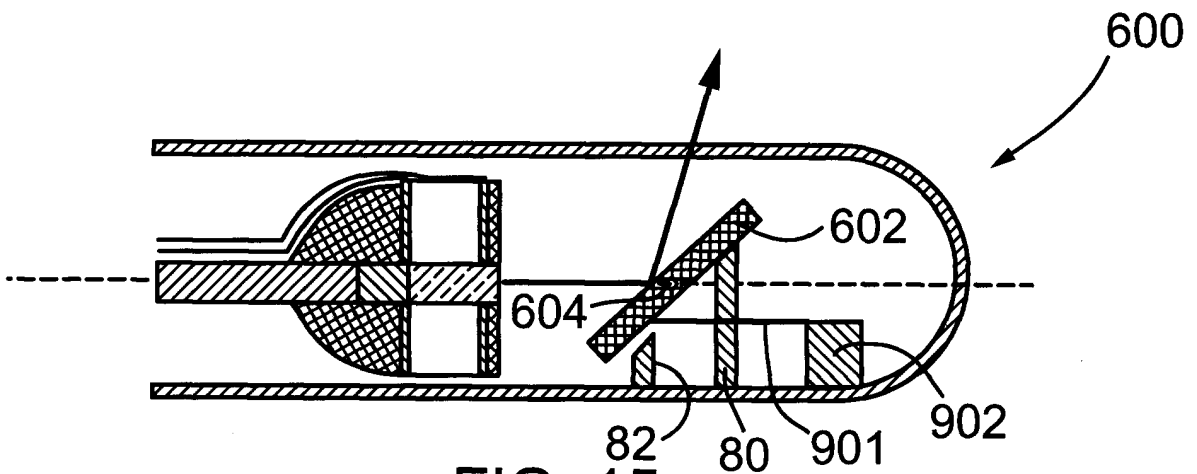
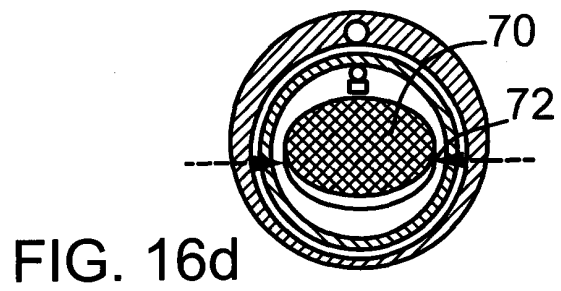
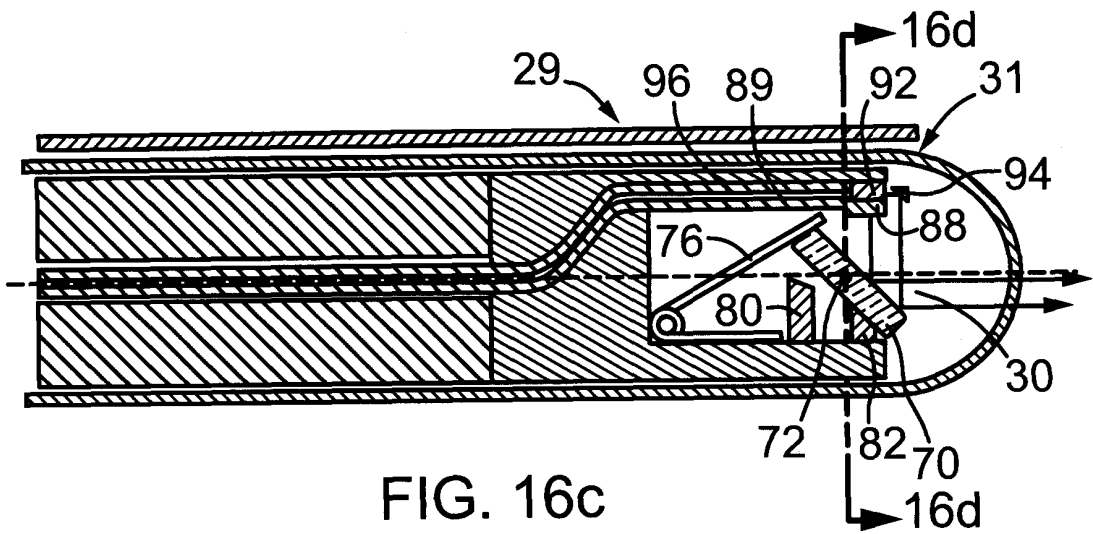
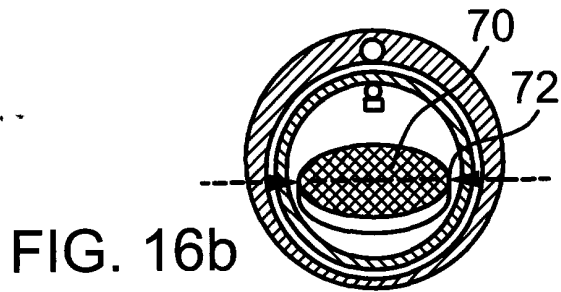
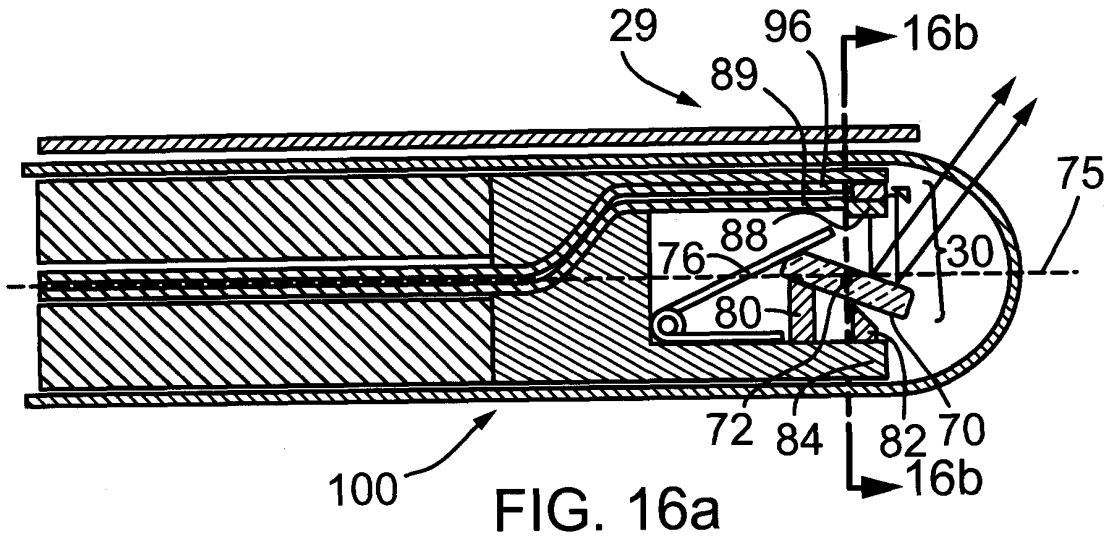


FIG. 15c



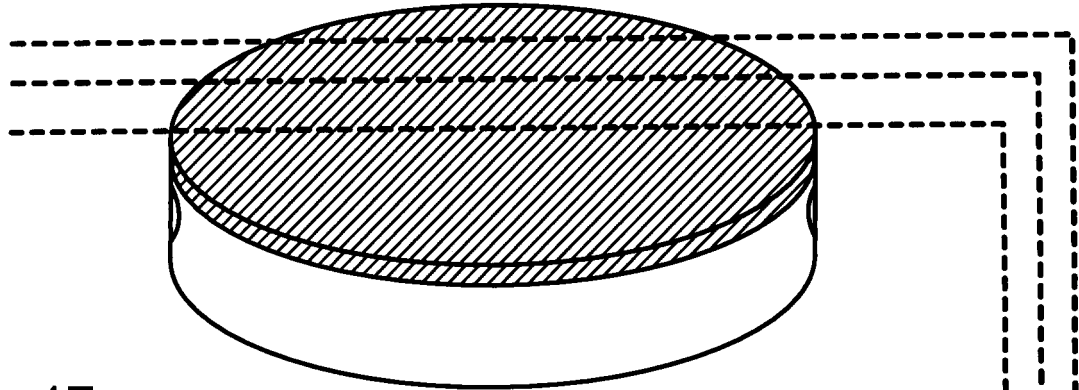


FIG. 17a

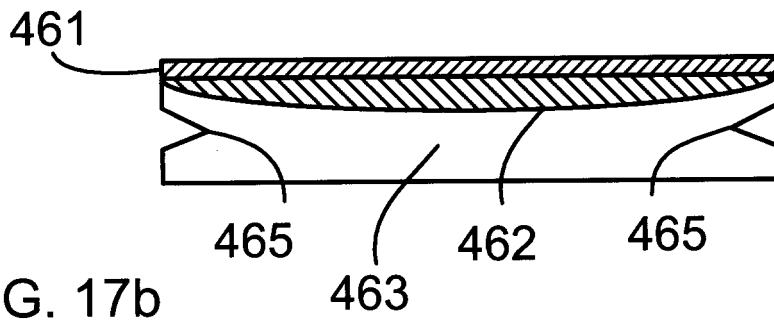


FIG. 17b



FIG. 17c

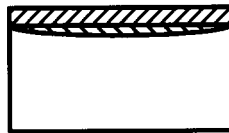


FIG. 17d

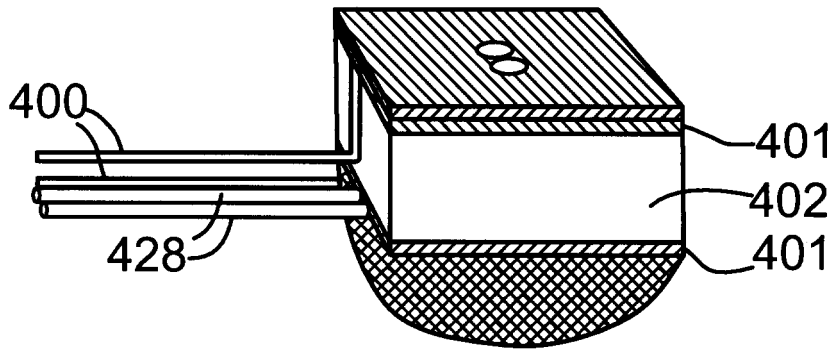


FIG. 18a

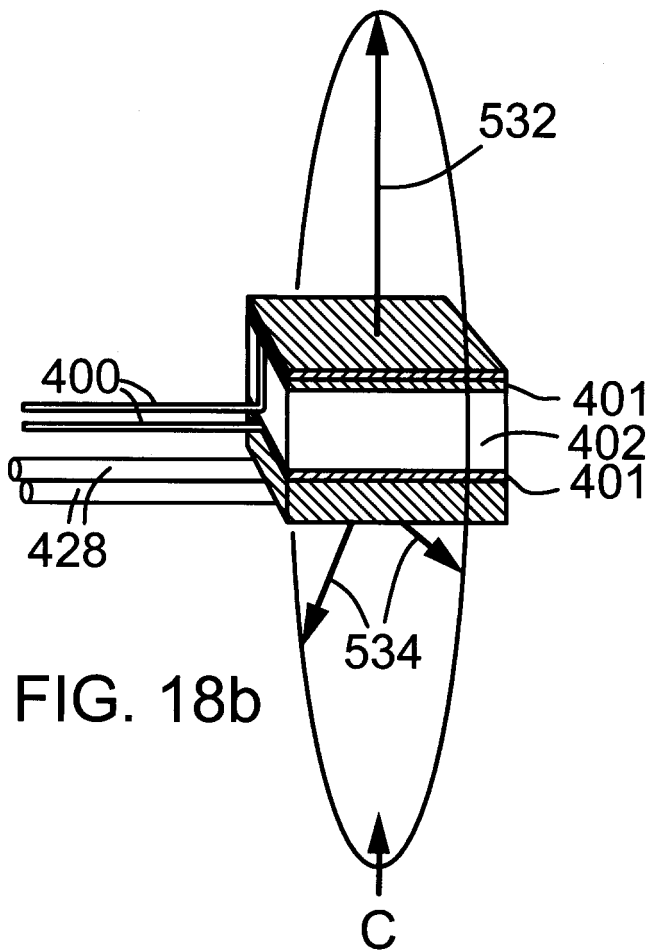


FIG. 18b

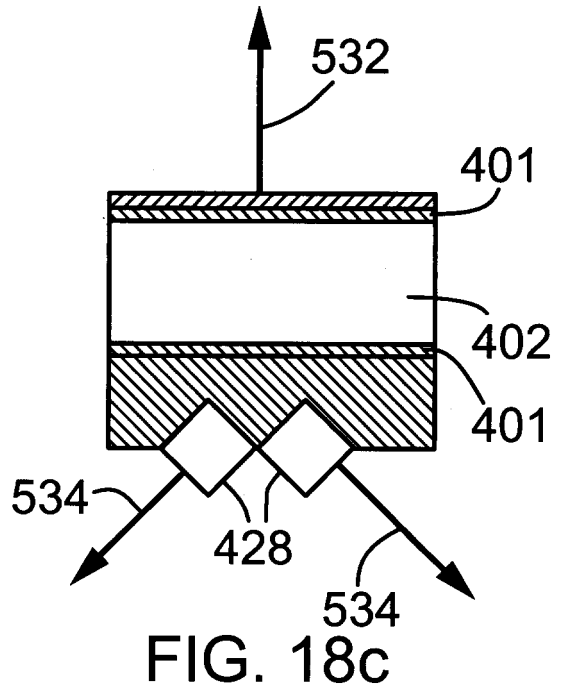


FIG. 18c

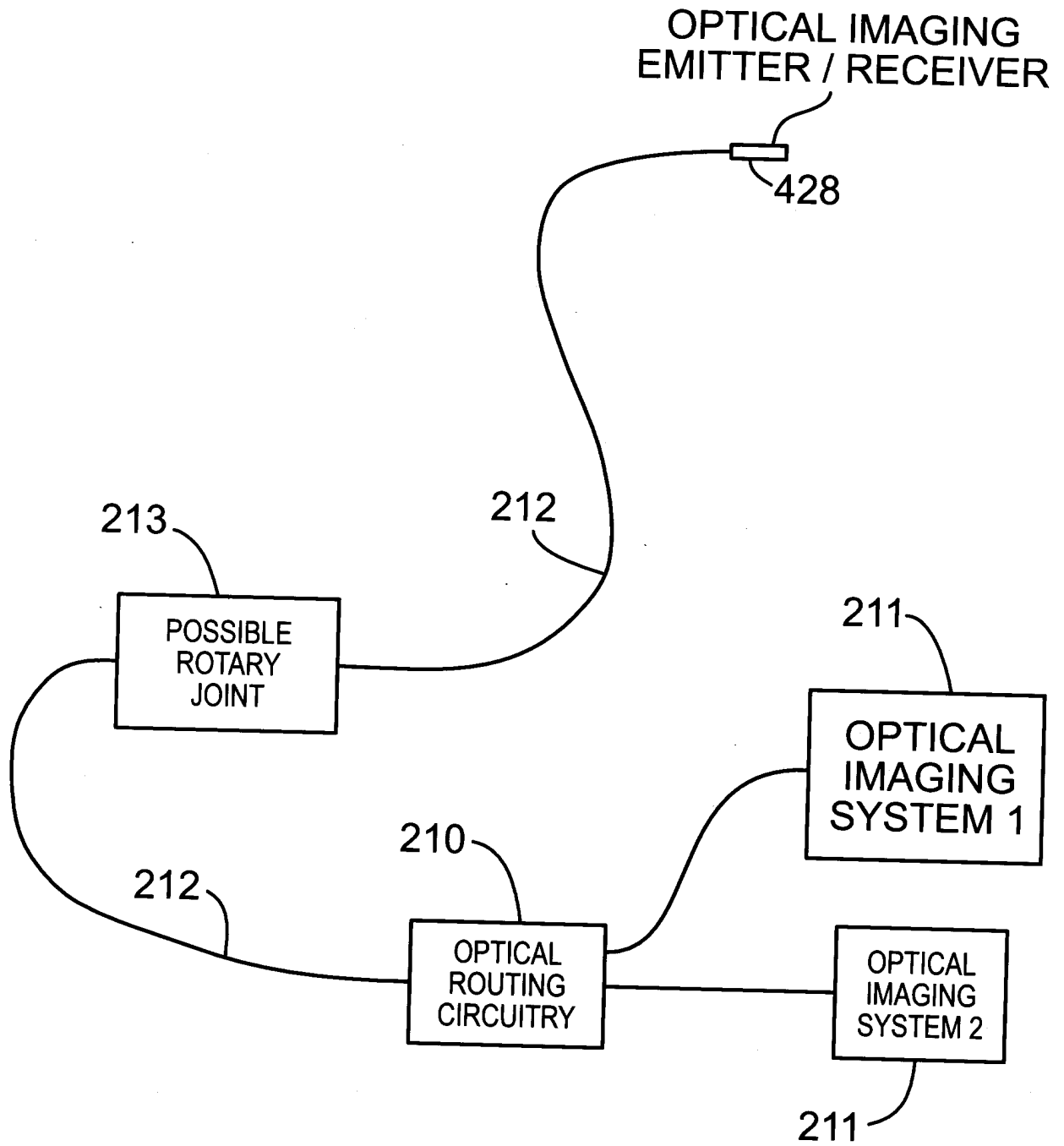


FIG. 19



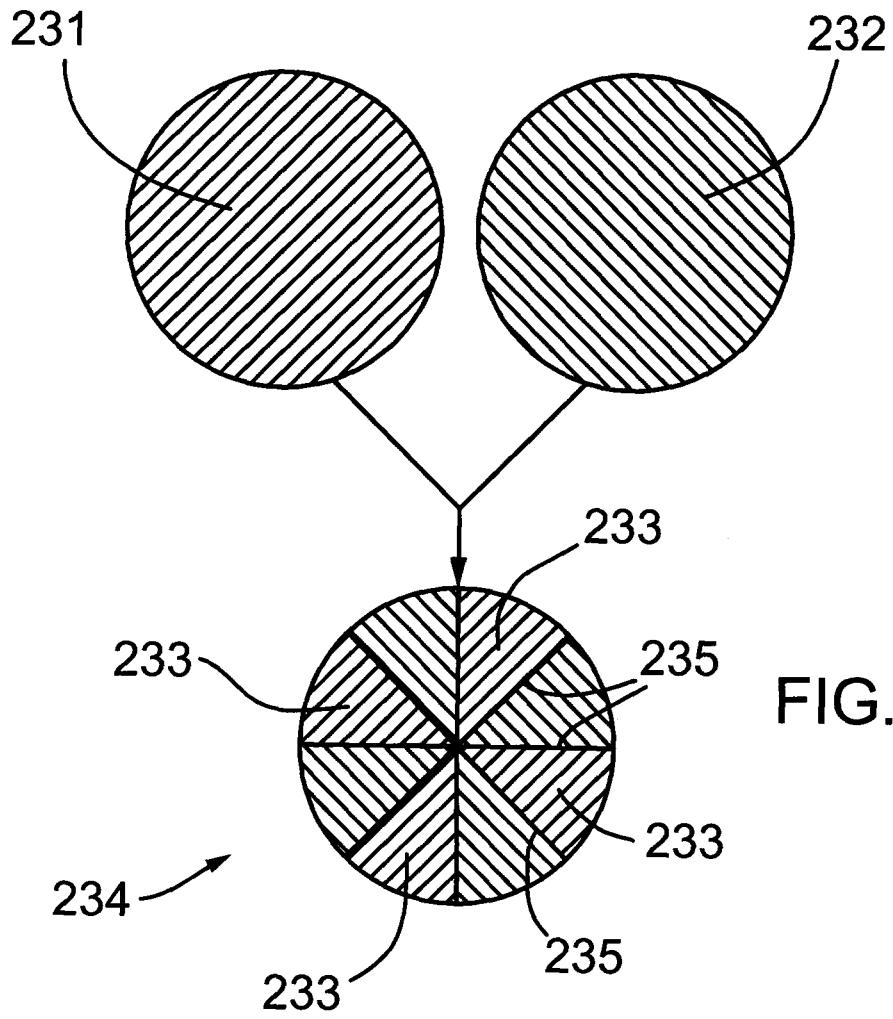


FIG. 20a

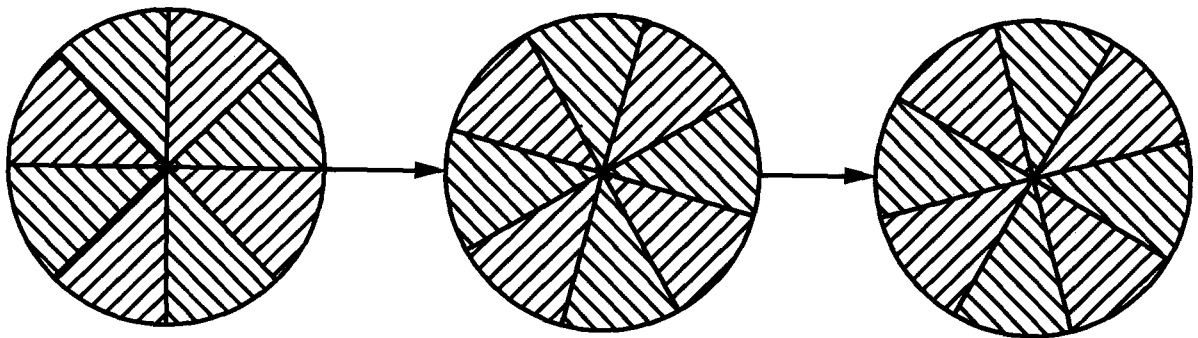


FIG. 20b

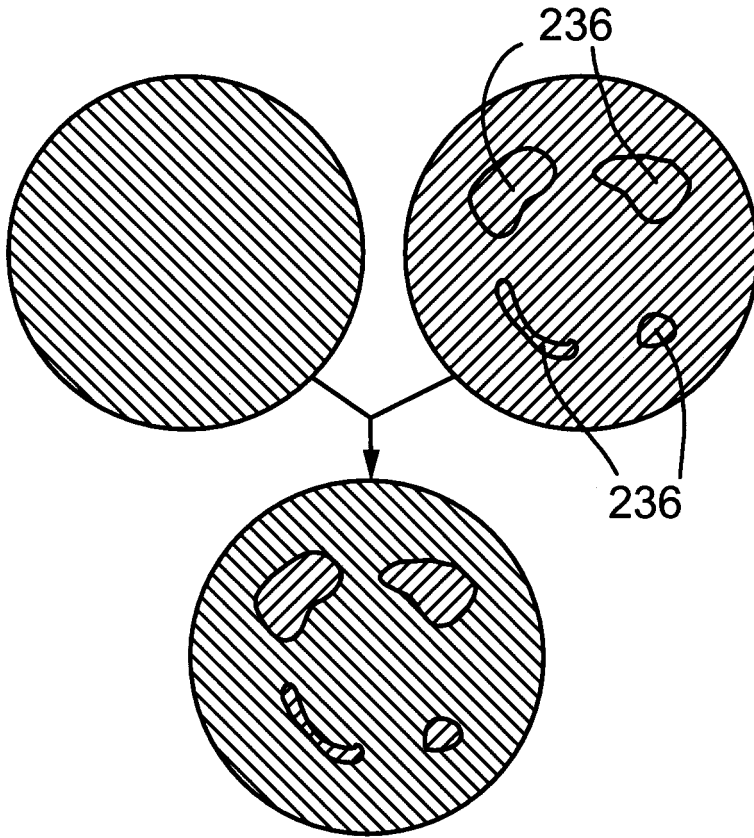


FIG. 21a

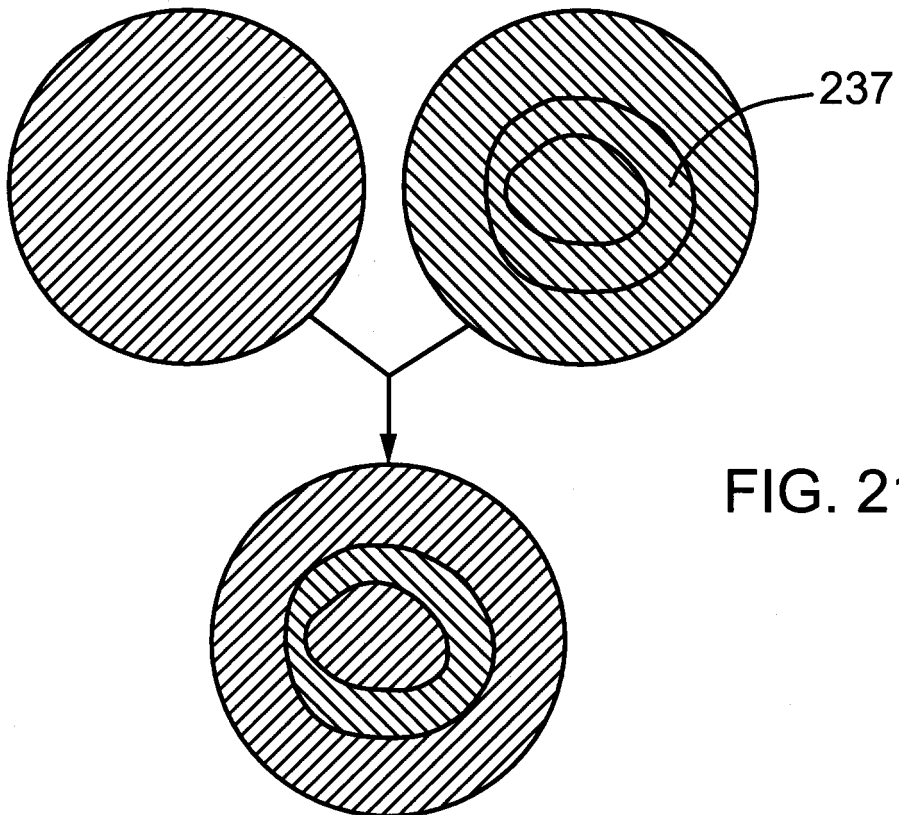


FIG. 21b

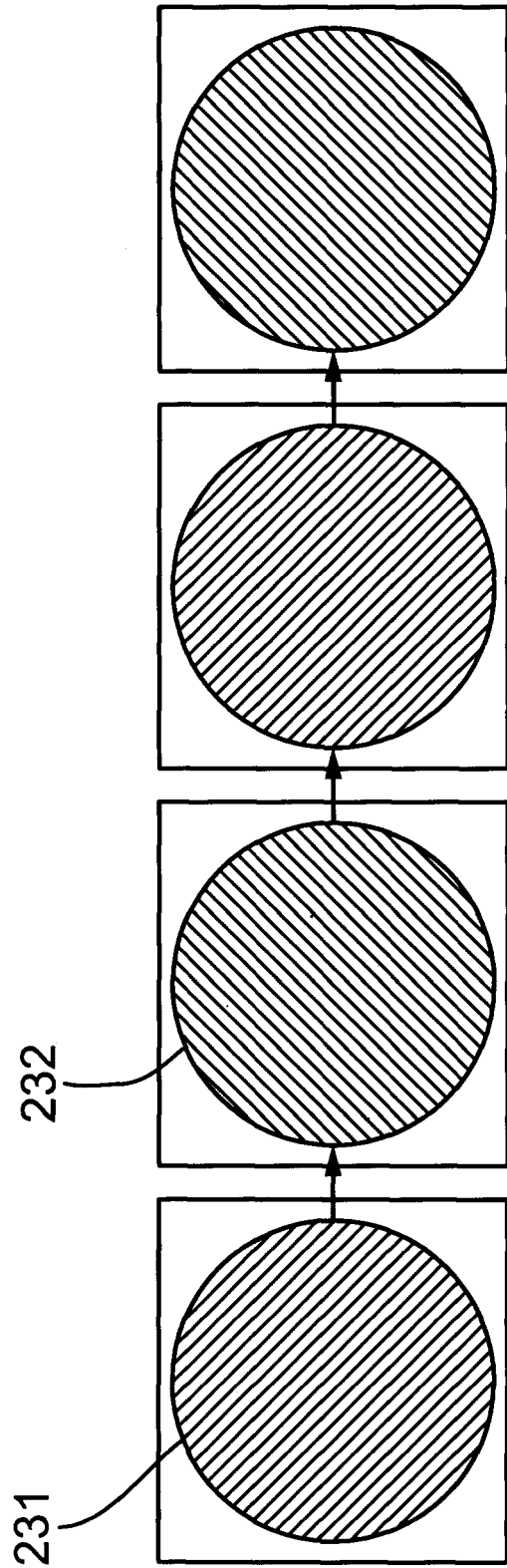
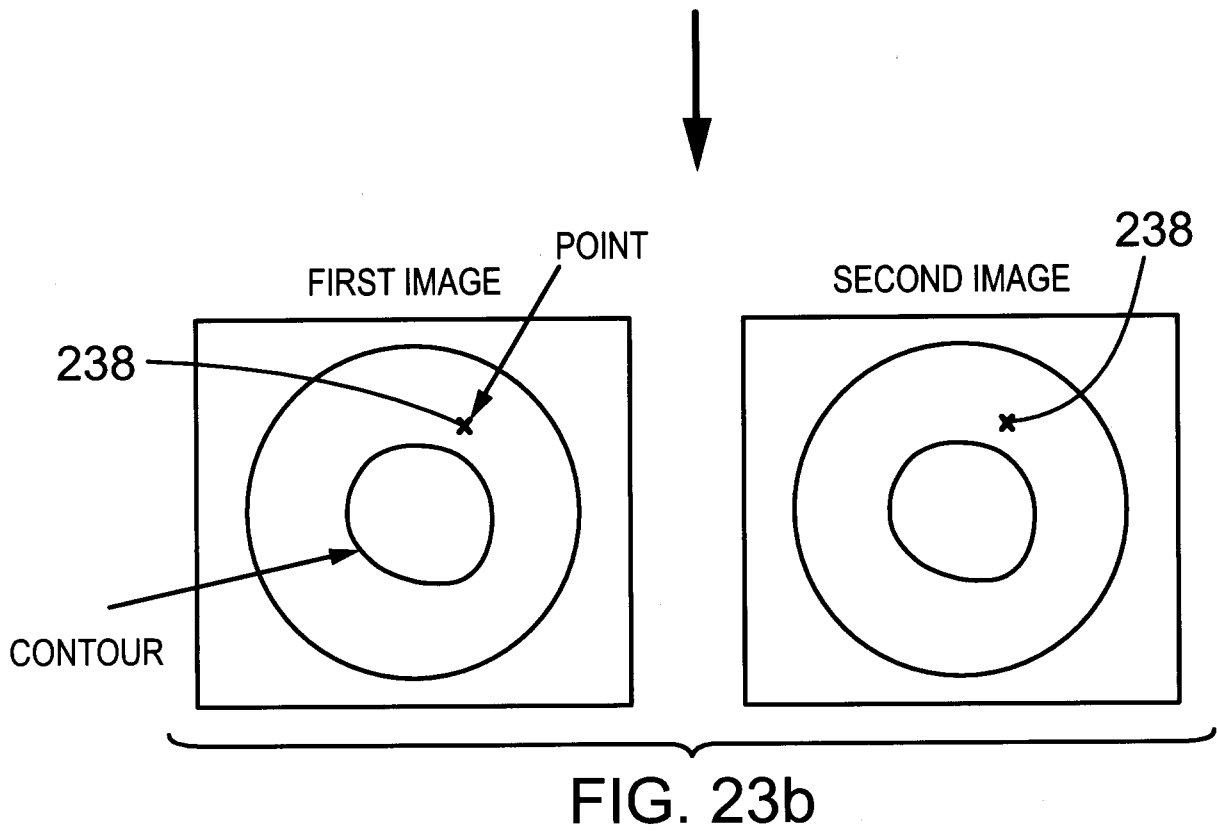
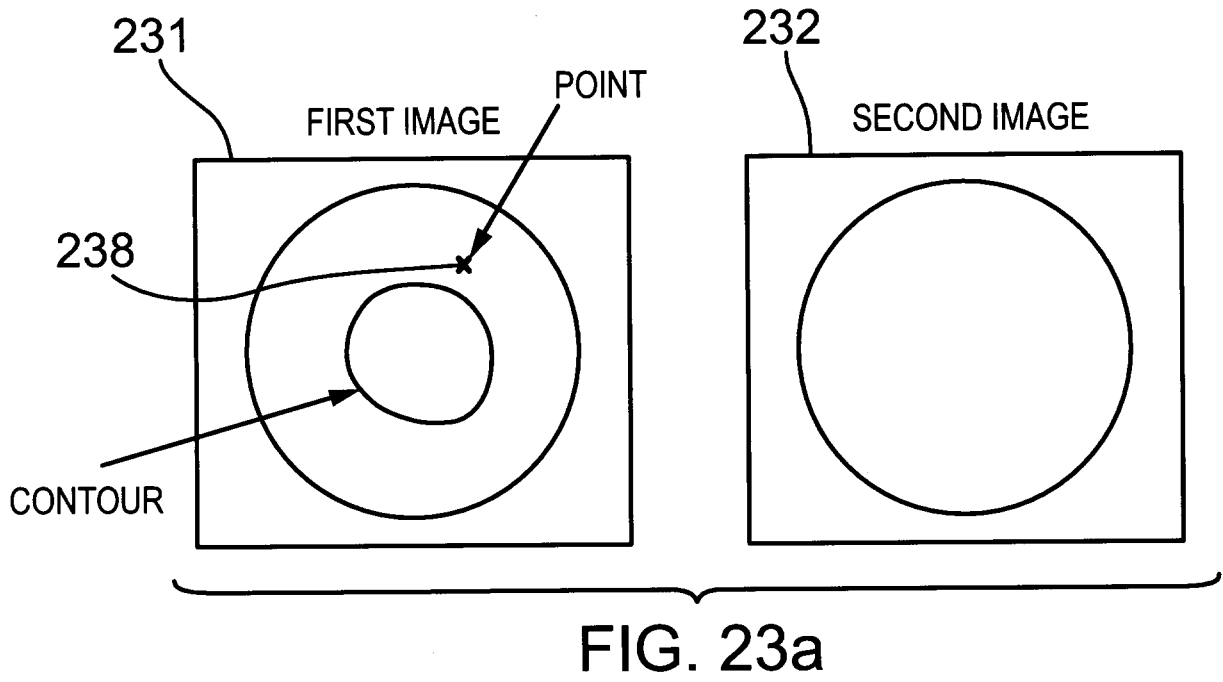
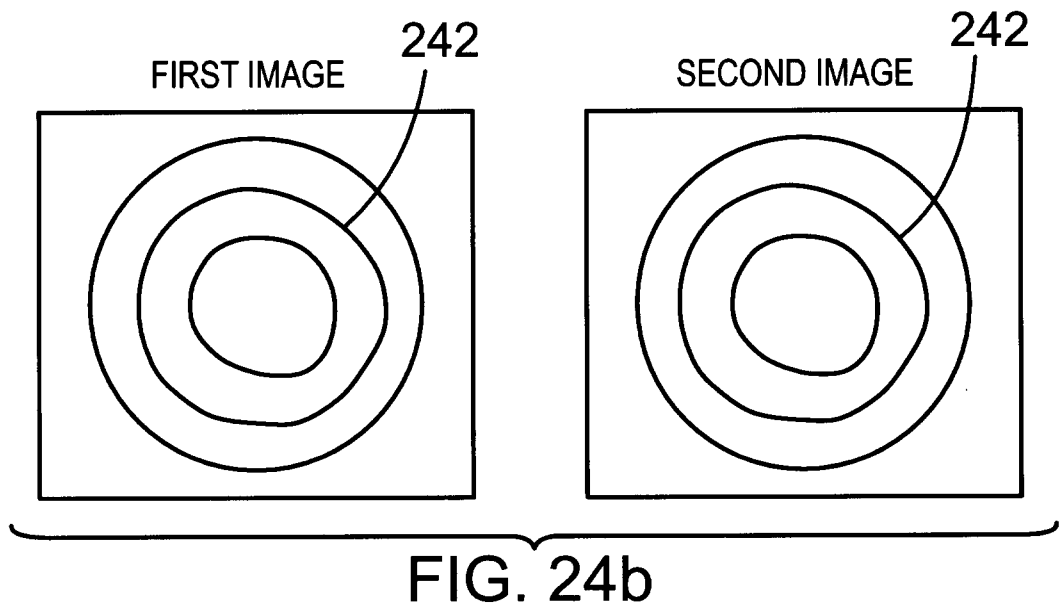
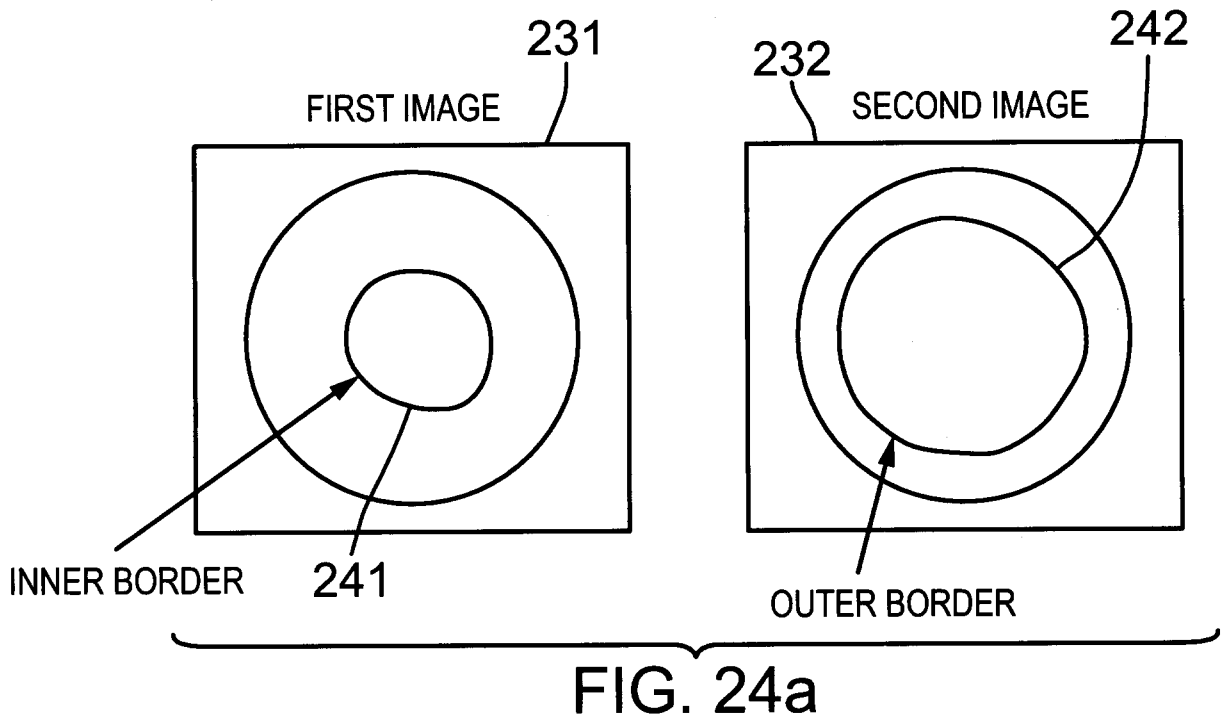


FIG. 22





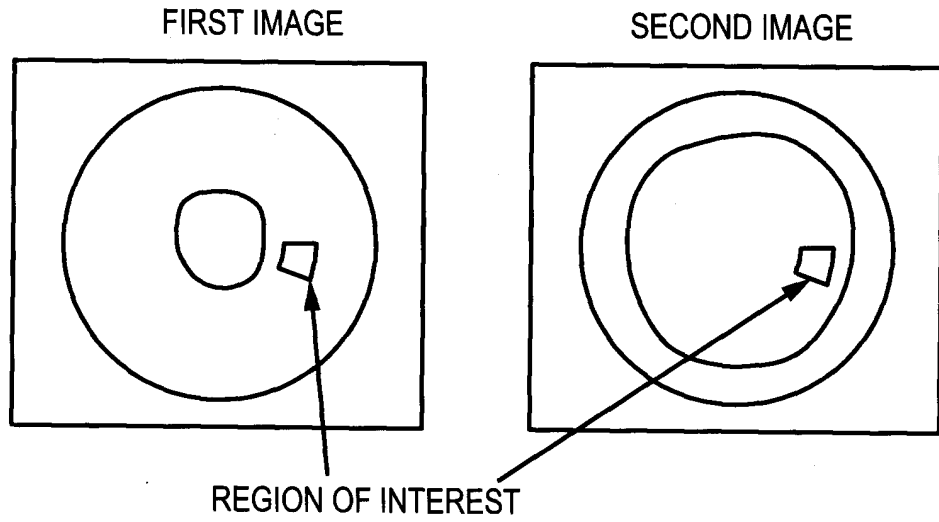


FIG. 25a

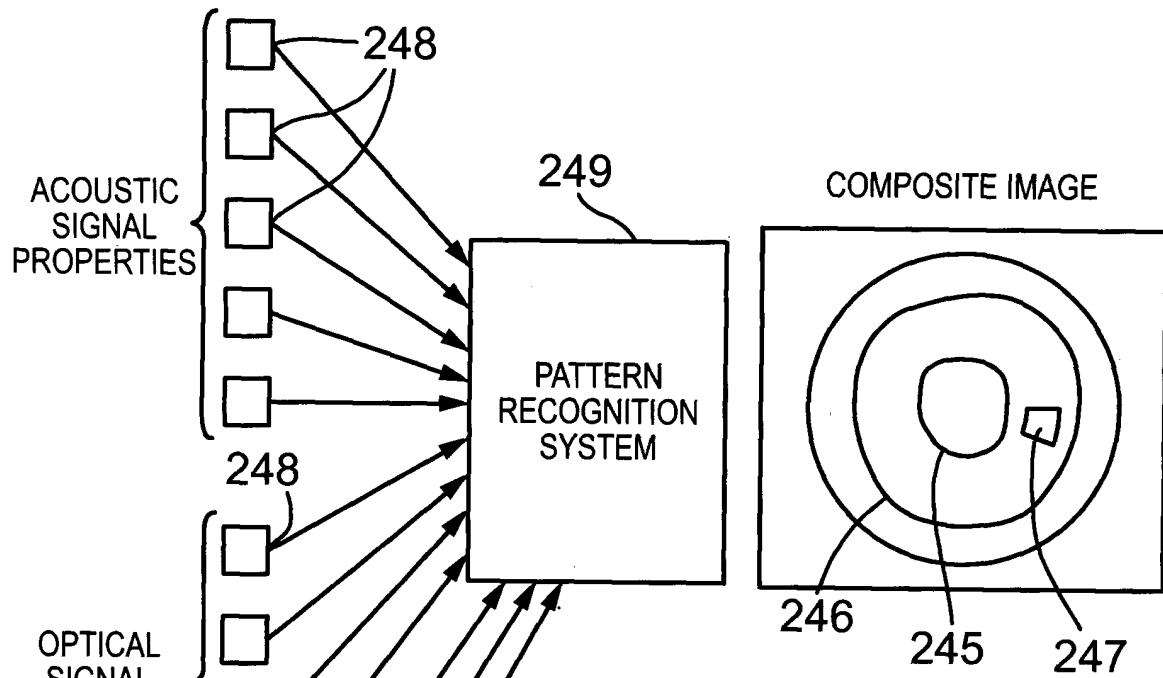


FIG. 25b

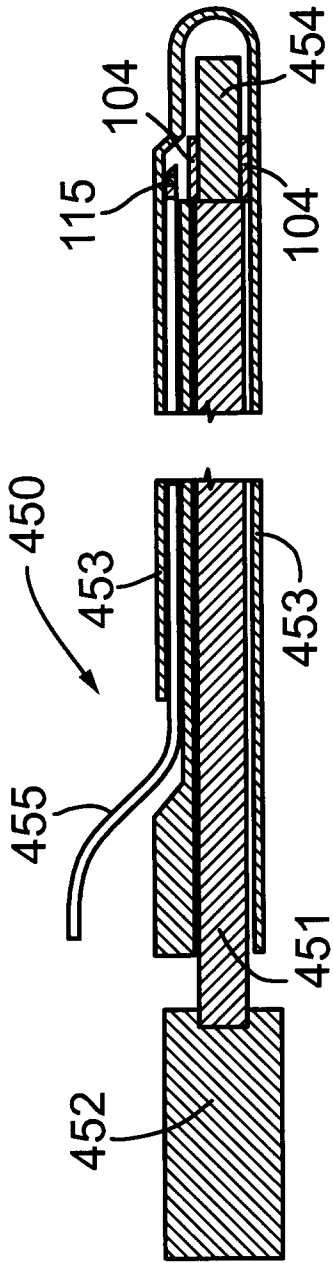


FIG. 26a

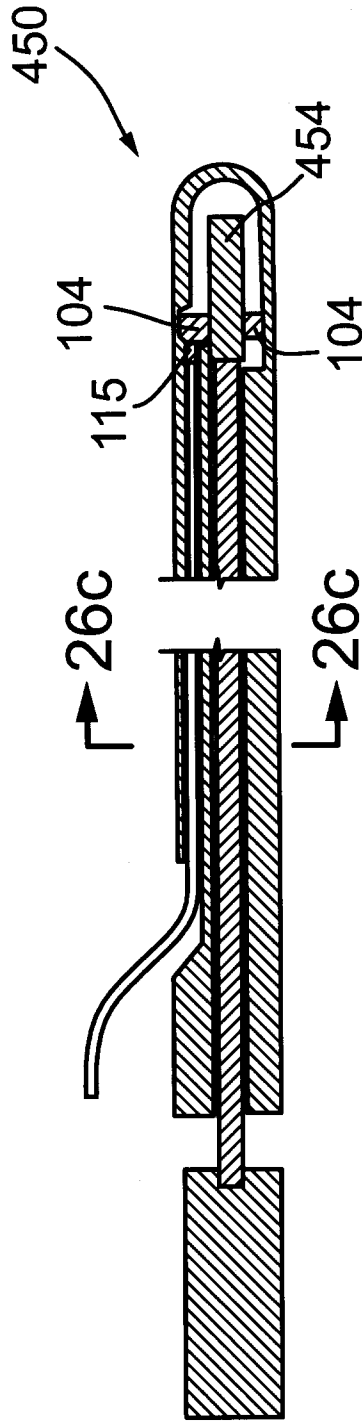


FIG. 26b

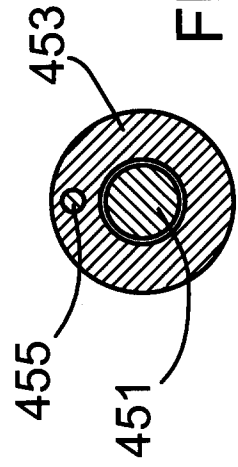


FIG. 26c

**REFERENCES CITED IN THE DESCRIPTION**

*This list of references cited by the applicant is for the reader's convenience only. It does not form part of the European patent document. Even though great care has been taken in compiling the references, errors or omissions cannot be excluded and the EPO disclaims all liability in this regard.*

**Patent documents cited in the description**

- US 7289842 B [0012] [0033]
- US 4794931 A [0013]
- US 5429136 A, Milo [0014]
- US 6110121 A, Lenker [0014]
- US 6592526 B [0014]
- US 5606975 A [0015]
- US 5651366 A [0015]
- US 4951677 A, Crowley [0016]
- US 6200268 B [0018] [0173]
- WO 2006121851 A, Hossack [0020]
- US 6134003 A, Tearney [0021]
- US 6485413 B [0022]
- US 6178346 B [0025]
- US 5718231 A [0026]
- US 20060116571 A [0032]
- US 291593 A [0032]
- US 415848 A, Park [0034]
- US 6390978 B1 [0045]

**Non-patent literature cited in the description**

- **DE KORTE et al.** *Circulation*, 09 April 2002, vol. 105 (14), 1627-30 [0018]
- **GOERTZ et al.** *Invest Radiol.*, August 2006, vol. 41 (8), 631-8 [0018]
- **MAO et al.** *Appl Opt.*, 10 August 2007, vol. 46 (23), 5887-94 [0023]
- **MOTZ et al.** *J Biomed Opt.*, March 2006, vol. 11 (2) [0028]
- **CAPLAN et al.** *J Am Coll Cardiol.*, 18 April 2006, vol. 47 (8), C92-6 [0028]
- *Radiology*, June 2004, vol. 231 (3), 659-66 [0028]
- *Journal of American College of Cardiology*, 04 September 2007, vol. 10 (50), 933-39 [0029]
- **BREZINSKI et al.** *Heart*, May 1997, vol. 77 (5), 397-403 [0029]
- **DANIELS ; PRICE.** *Ultrasound in Medicine and Biology*, 1991, vol. 17 (3), 297-308 [0152]
- **KLINGENSMITH.** *IEEE Transactions on Medical Imaging*, 2000, vol. 19, 652-662 [0168]
- **NAIR et al.** *Circulation*, 2002, vol. 106 (17), 2200-2206 [0173]



专利名称(译)	成像探头与组合超声和光学成像手段		
公开(公告)号	<a href="#">EP2111165A4</a>	公开(公告)日	2013-10-16
申请号	EP2008706237	申请日	2008-01-21
[标]申请(专利权)人(译)	桑尼布鲁克健康科学中心		
申请(专利权)人(译)	Sunnybrook健康科学中心		
当前申请(专利权)人(译)	Sunnybrook健康科学中心		
[标]发明人	COURTNEY BRIAN MUNCE NIGEL ROBERT THIND AMANDEEP SINGH YANG VICTOR XIAO DONG FOSTER FRANCIS STUART		
发明人	COURTNEY, BRIAN MUNCE, NIGEL, ROBERT THIND, AMANDEEP SINGH YANG, VICTOR, XIAO DONG FOSTER, FRANCIS, STUART		
IPC分类号	A61B8/12 A61B1/04 A61B1/07 G01D5/347		
CPC分类号	A61B1/00172 A61B1/00183 A61B5/0035 A61B5/0062 A61B5/0066 A61B5/0084 A61B5/0095 A61B5/6852 A61B5/7207 A61B5/742 A61B8/12 A61B8/4281 A61B8/445 A61B8/4461 A61B8/483 A61B8/5238 A61B8/5292 A61B8/543 A61B2562/0233 G10K11/002 A61B1/00112 G02B6/3604 G02B23/2423		
优先权	60/881169 2007-01-19 US		
其他公开文献	EP2111165B8 EP2111165B1 EP2111165A1		
外部链接	<a href="#">Espacenet</a>		

#### 摘要(译)

本发明提供了用于成像探针的扫描机制，该探针用于使用高分辨率成像对哺乳动物组织和结构进行成像，包括高频超声和/或光学相干断层扫描。成像探针包括可调节的旋转驱动机构，用于将旋转运动传递给成像组件，该成像组件包含向周围区域发射能量的光学或超声换能器。该成像组件包括扫描机构，该扫描机构包括可移动构件，该可移动构件被配置为沿着相对于所述纵向轴线可变的角速度沿着所述细长空心轴的路径传递能量束，以给出成像组件的前向和侧向观察能力。可移动构件以使得可变速角速度是成像组件的角速度的函数的方式安装。