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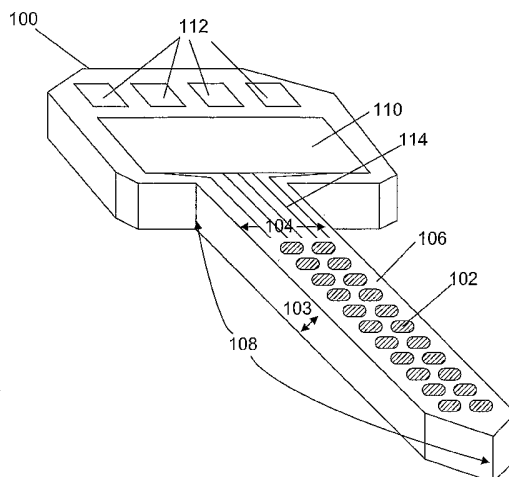
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(54) Title: IMPLANTABLE ULTRASONIC TRANSDUCER SYSTEMS AND METHODS



(57) Abstract: A miniaturized micromachined ultrasonic transducer 100 is disclosed herein for implantation in tissue to achieve high-resolution images. An array of ultrasonic transducers 102 is fabricated in a carrier 700 using integrated circuits (IC) technology. The carrier 700 is micromachined into a needle shape with a cross-section small enough for the array of ultrasonic traducers 102 to be inserted into tissue for high- resolution medical imaging. The length of the probe shank portion 106 of the transducer 100 is dependent on the quantity of transducers 102 to be arranged in the array. Compared with commercial intravascular ultrasonic devices, this micromachined ultrasonic imager is small enough to allow monolithic integration of a flexible silicon ribbon cable 504 for electrical interconnection from the array of transducers to external electronic. With the silicon ribbon cable 504 as the connector for the array of transducers 102, a catheter is not necessary.

WO 2005/114820 A2

Implantable Ultrasonic Transducer Systems and Methods

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CROSS-REFERENCE TO RELATED APPLICATION

[0001] This application claims priority to copending U.S. provisional application entitled, "Implantable Ultrasonic Transducer Systems and Methods," having ser. no. 60/571,179, filed May 14, 2004, which is entirely incorporated herein by reference.

TECHNICAL FIELD

[0002] The present disclosure is generally related to ultrasonic imaging and, more particularly, is related to ultrasonic transducer systems that can be implanted inside a body and related methods.

BACKGROUND

[0003] Traditional ultrasonic systems are bulky and are typically operated outside the body. This prevents them from acquiring high-resolution images for anatomical structures deep inside the tissue. Commercial intravascular ultrasound (IVUS) can be inserted through a blood vessel for examining vessels in the body and the arteries in the heart. However, its size (3-4 millimeters in diameter) limits its use to large vessels.

[0004] Evolved from sonar technology, ultrasonic imaging has become a popular tool for medical diagnosis since the early 1970s. In this imaging process, high frequency sound waves are used to interact with the human body so that anatomical structures,

blood flow velocity and other diagnostic information can be obtained. Ultrasound is largely considered the second most widely used diagnostic imaging modality (next to X-ray). Gray scale (B-mode) ultrasound became popular in the 1970s. Color Doppler became clinically useful in the 1980s. In the 1990s, all high-end systems are fully digital. The next generation systems will continue to take advantage of advanced signal processing techniques, faster clock rates, larger memory capacity and higher circuit density. The advantages of ultrasonic imaging systems include non-invasive image formation of anatomical structures, non-invasive detection of moving objects, real-time acquisition, portability and cost. Current clinical applications include OB/GYN, vascular, cardiac, transcranial, abdominal, musculoskeletal, endo-vaginal, endo-rectal, ocular, intra-vascular, intra-operative, among others. With the recent advancement of electronic technologies and more understanding of imaging mechanisms, the clinical capabilities of ultrasonic systems have been expanding rapidly.

[0005] Traditionally, piezoelectric thin-films have been the most widely used building material for ultrasonic transducers. These piezoelectric ultrasonic transducers typically operate at voltages higher than 100 volts and are normally arranged in one-dimensional (1-D) arrays. With the increasing demand for real-time three-dimensional (3-D) images in medical diagnosis applications, many efforts have been attempted to build a two-dimensional (2-D) piezoelectric ultrasonic array. Although initial success has been achieved and some commercial devices have been introduced, several technical obstacles such as electrical impedance matching and interconnections still challenge the engineers working in this field. Another problem associated with piezoelectric ultrasonic transducers is that the acoustic impedance of many piezoelectric thin-films does not match that of the fluids encountered in most

applications. The problem of impedance mismatch worsens in gaseous fluids. For example, the acoustic impedance of piezoceramics, water, and air are $30 \times 10^6 \text{ kg/m}^2\text{s}$, $1 \times 10^6 \text{ kg/m}^2\text{s}$ and $400 \text{ kg/m}^2\text{s}$, respectively. Under such significant impedance mismatch, most of the ultrasonic waves will bounce back at the transducer-fluids interface. Although some approaches, like adding impedance matching layers, have been used to alleviate the problem, poor efficiency is still a fundamental limitation of piezoelectric ultrasonic systems used in fluids.

[0006] As limits including efficiency, high operation voltage, bandwidth, surface displacement, and electrical interconnect/impedance-match of piezoelectric ultrasonic systems pose constraints in system design, alternatives have been presented. Since 1989, there have been many efforts in applying micromachining technology in capacitive ultrasonic transducer implementations. A capacitive ultrasonic transducer comprises a suspended membrane and a counter electrode fixed to a supporting substrate. When an a/c (alternating current) signal is applied between the membrane and the fixed electrode, the alternating Coulomb forces drive the membrane to vibrate and emit ultrasonic waves. At the same time, upon reception of impinging ultrasonic waves, the membrane will deform and change the capacitance. This change of capacitance is converted into electrical signals, which are used to interpret the ultrasonic waves. As implemented with the technology for fabricating integrated circuits, these micromachined ultrasonic elements are readily arranged into one-dimensional and two-dimensional arrays and are compatible with integration with front-end electronics. The addition of front-end complimentary metal oxide semiconductor (CMOS) circuitry improves the signal-to-noise ratio and enhances system reliability. Different from piezoelectric ultrasonic devices which normally require a thick piezoelectric film up to several millimeters for high frequency

operation, the micromachined ultrasonic devices are made of thin films which are typically of a thickness ranging from submicron to a few microns. No matter whether it is a piezoelectric or capacitive ultrasonic system, one of the fundamental limitations of these traditional ultrasonic medical imagers is that they are bulky and do not fit well with implantation applications. Ultrasonic imaging near the surface of the body is capable of resolutions less than a millimeter. The resolution decreases with the depth of penetration since longer wavelengths are used (the attenuation of the waves in tissue goes up with increasing frequency). The use of longer wavelengths implies lower resolution since the resolution limit of any imaging process is proportional to the wavelength of the imaging wave. This is a fundamental limit of many imaging systems. In many medical diagnosis applications, for example, imaging of anatomical structures of a liver, it is desired that high-resolution images be obtained for structures deep inside the tissue. High-resolution imaging requires use of short-wavelength ultrasonic waves.

[0007] An intravascular ultrasound system (IVUS) can be inserted into a blood vessel for diagnosis. Intravascular ultrasound uses a long, flexible tube called a catheter, about 3 to 4 millimeters in diameter, to accommodate a probe at its tip from which the ultrasonic signals are generated and received. The catheter is inserted into the patient's artery or vein through a small incision in her/his skin. The position of the ultrasonic probe inside the body is traced by X-ray and the ultrasonic images are recorded by signal-processing electronics connected at the other end of the catheter. With its unique capability, intravascular ultrasound is able to identify problems that other tests cannot adequately reveal and has become a popular tool for diagnosing vessel blockage or abnormalities. A limit of IVUS comes from its size. A 3 to 4 millimeters diameter catheter can only be inserted through large blood vessels. It is

too large to reach fine vessels and organs. However, in many medical diagnoses it is important to acquire image information from the fine vessels and the anatomical structures deep inside an organ.

[0008] Traditional integrated circuits reside on a bulk piece of silicon which is typically several hundred microns thick. Of these several hundred microns only the top one or two microns are structures built for the functional devices (e.g., transistors), the rest of the bulk silicon serving as a mechanical support such that the silicon wafer does not break during integrated circuits fabrication. Similarly, traditional piezoelectric ultrasonic devices and MEMS (Micro-Electro-Mechanical-System) ultrasonic systems use bulky substrates, which are normally silicon, glass, sapphire, or plastic to accommodate the transducers. In the case of piezoelectric ultrasonic devices, the thickness of the piezoelectric film alone can be as thick as several millimeters for some high frequency transmitters.

[0009] With the advent of micromachining technology, the substrate of micro-transducers can be shaped to micron or sub-micron scale. With proper packaging, such micro transducer chips can be inserted in tissue for interfacing with cells or neurons or implanted in the heart for monitoring the blood pressure and flow. One example is the silicon neural probe developed at the University of Michigan. Researchers at the University of Michigan used selective boron diffusion to define an area on a silicon substrate over which thin-film electrodes and CMOS interface circuits were integrated for recording electro-chemical voltages from neurons and for electrical stimulation. After completion of the fabrication process for sensors and electronics, the silicon substrate was dissolved in ethylenediamine pyrocatechol (EDP), leaving only a 16 μ m-thick boron doped area which worked as a mechanical support for the sensors and transistors. Defined by photolithography, a deep boron

diffusion and a dry etching, the substrate of such silicon neural probes can be shaped to any geometrical configuration and any thickness ranging from 0.1 micron to tenths of microns. A typical thickness of these silicon neural probes is 16 microns.

Generally the thickness is tailored to fulfill mechanical strength requirements for the application. Similar technology has been applied to implementation of implantable micro-sensors for cardiovascular monitoring.

SUMMARY

[0010] Embodiments of the present disclosure provide a system and method for implantable ultrasonic transducers.

[0011] Briefly described, in architecture, one embodiment of the system, among others, can be implemented with a micromachined transducer array and a micromachined cable that is monolithically integrated with or bonded to the micromachined transducer array. Another embodiment can be implemented with a substrate, a dielectric film adjacent to the substrate, a thin film electrode adjacent to the dielectric film, and a thin film membrane in contact with the dielectric film and separated from the thin film electrode by a gap. Embodiments of the present disclosure can also be viewed as providing methods for fabricating a micromachined ultrasonic array. In this regard, one embodiment of such a method, among others, can be broadly summarized by the following steps: providing a substrate; and fabricating a micromachined ultrasonic array on the substrate.

[0012] Other systems, methods, features, and advantages of the present disclosure will be or become apparent to one with skill in the art upon examination of the following drawings and detailed description. It is intended that all such additional systems,

methods, features, and advantages be included within this description, be within the scope of the present disclosure, and be protected by the accompanying claims.

BRIEF DESCRIPTION OF THE DRAWINGS

[0013] Many aspects of the disclosure can be better understood with reference to the following drawings. The components in the drawings are not necessarily to scale, emphasis instead being placed upon clearly illustrating the principles of the present disclosure. Moreover, in the drawings, like reference numerals designate corresponding parts throughout the several views.

[0014] FIG. 1 is a perspective view of an exemplary embodiment of an implantable ultrasonic transducer.

[0015] FIG. 2 is a view of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 1.

[0016] FIG. 3 is a perspective view of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 1 with microheaters.

[0017] FIG. 4 is a view of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 3.

[0018] FIG. 5 is a perspective view of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 1 with a ribbon cable connection.

[0019] FIG. 6 is a view of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 5.

[0020] FIGs. 7A-7J are views of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 1 at progressive stages of fabrication.

[0021] FIGs. 8A-8J are views of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 5 at progressive stages of fabrication.

[0022] FIGs. 9A and 9B are views of a cross section of an exemplary embodiment of the implantable ultrasonic transducer of FIG. 5 as it is separated from the substrate.

DETAILED DESCRIPTION

[0023] Embodiments of a micromachined ultrasonic imager array are disclosed that may be smaller than a human hair and can be implanted into tissue and fine vessels for medical imaging. A flexible yet robust silicon ribbon cable is also disclosed for integration with the ultrasonic imager array as the electrical connection to external electronics such that no bulky catheter is needed for this implantable system. With this arrangement, the cross-sectional dimensions of all the components to be implanted into tissue, including the sensors and the interconnects, are small such that the disruption to the tissue due to insertion of the imaging tool is minimized. Addition of microheaters and electrodes on the imager probe further expand the capability of this tool to hyperthermia treatment and electrical stimulation of cells and tissue. Exemplary embodiments of ultrasonic systems that are disclosed enable high-resolution imaging of tissues, organs, and cells and treatment of tumors deep inside the tissue.

[0024] Such miniature ultrasonic systems provide a unique tool for acquiring high resolution images of anatomical structures deep inside an organ/tumor or fine vessels, as well as larger vessels. Applications of such systems include high-resolution acoustic biopsy (for example, one could insert such an ultrasonic array into the liver to check the fibrosis grade or fatty change of liver) and in-situ monitoring of blood flow. The disclosed systems can also be used in neural prosthesis and imaging of neurons.

[0025] Exemplary embodiments disclosed herein include a miniaturized micromachined ultrasonic transducer that can be implanted in tissue for high

resolution imaging applications. FIG. 1 demonstrates an ultrasonic transducer 100 fabricated using integrated circuits (IC) technology. The carrier may be micromachined into a needle shape with a cross-section small enough to be inserted into tissue for high-resolution medical imaging. The length of the probe can vary depending on the quantity of the transducers in the array.

Compared with commercial intravascular ultrasonic devices, an exemplary embodiment of a micromachined ultrasonic imager allows monolithic integration of a flexible silicon ribbon cable for electrical interconnection from the transducer array 100 to the external electronics. With a silicon ribbon cable as the connector for the imager array, a catheter is no longer needed.

[0026] FIG. 1 shows a perspective of an exemplary embodiment of an implantable ultrasonic imager array with a two-row array of MEMS ultrasonic transducers 102 arranged on a micromechanical substrate 106. Although a 2-row array is illustrated, any 1-D or 2-D array can be arranged on the micromachined substrate 106. The substrate of exemplary embodiments of an imager array can be silicon, glass, sapphire or any material that is compatible with the thin-film processes used to build the ultrasonic transducers 102. The thickness 103 of the micromachined substrate (or the probe shank) 106 can be tailored depending on the requirement of mechanical strength for the application. The width 104 of probe shank 106 can be designed depending on the size of the transducers 102 and how many rows of transducers 102 are to be arranged on probe shank 106.

[0027] The length 108 of the probe shank 106 varies depending on the size of the imager array. The transducer 102 includes a suspended membrane and a fixed electrode. In order to avoid direct exposure of the tissue/cells to high electrical fields, the suspended membrane is connected to the electrical ground while the fixed

electrode is connected to the electrical signal line. As the membrane is made of highly conductive material, for example, highly doped polysilicon, the shielding effect will constrain the electric field in the region under the membrane. Similar arrangements may also be made for the electrical interconnect.

[0028] The thin film constituting the fixed electrodes of transducers 102 may also be used to form the electrical interconnects 114 from the ultrasonic elements to the CMOS circuits 110 at the base region. These electrical interconnects 114 are passivated by the same layer of conducting thin film used to build the membrane. This forms a coaxial-cable like structure. This structure not only confines the electrical fields from reaching the tissue but also reduces the noise pickup from tissue fluids. In order to insulate the membrane from the tissue fluids and to protect it from tissue fluid attack, a passivation layer is added on top of the membrane. Normally this passivation structure comprises dielectrics and a thin layer of parylene C. Bonding pads 112 are used to connect to monitoring equipment (not pictured).

[0029] As illustrated in FIG. 2, drum structures 200 made of doped polysilicon or silicide are used when emitting and sensing ultrasonic waves. The drum structure 200 includes electrode 202 and suspended membrane 206, which define space 204. The space 204 between the suspended membrane 206 and the counter electrode 202 is vacuum-sealed so the tissue fluids will not flow into this region during operation. By applying an a/c electrical signal with proper bias on the lower polysilicon/silicide electrode 202, the membrane 206 will vibrate and generate ultrasonic waves. The driving voltage of the ultrasonic transducer may be low (<20 volts) and may be further reduced to enable driving transducers using conventional low-voltage CMOS transistors. Adding on-chip CMOS transmit-and-receive circuitry can simplify board level system design and reduce system cost. After these ultrasonic elements are

formed on a regular silicon wafer, the substrate can be micromachined using deep silicon etching from both sides of the substrate. Elements with different resonant frequencies can be hybrid in one array for some special imaging applications.

[0030] FIG. 3 illustrates an exemplary embodiment including microheaters with temperature sensor 302 and electrode 304 added on the same carrier with the ultrasonic transducers 102 for hyperthermia treatment and electrical stimulation of tissue, cell, or tumor. In addition to ultrasonic transducers 102, microheaters made of doped polysilicon or silicide and electrodes can also be added on the same carrier for hyperthermia treatment and electrical stimulation. In order to minimize the heat conduction from a heater to the whole silicon substrate, the heater 402 may be located on a dielectric window 404, as shown in FIG. 4, suspended on a hole created in the silicon substrate. This hole can be formed by selective boron diffusion and a wet etch in ethylenediamine pyrocatechol or by a masked dry etching. The electrode 304 can be formed from a layer of iridium oxide 408 (or other conducting material) deposited over dielectric 410. An electrode 304 may be made of gold, iridium, iridium oxide, or platinum, among other materials, which are connected to the external electronics through thin-film interconnects 114 (from FIG. 1). These thin-film interconnects 114 can be made of doped polysilicon, silicon carbide, polymers or any other conducting materials and are passivated with dielectrics so that they are electrically insulated from the ambient.

[0031] As shown in FIG. 5, an imager array 502 can be integrated monolithically with a silicon ribbon cable 504 for electrical connection to the external electronics. The silicon ribbon cable 504 can be made, in one embodiment, of single crystal silicon on which thin film interconnects are used to convey electrical signals between the transducers and the external electronics. The silicon ribbon cable 504 can be flexible,

yet strong and reliable and can replace the role of the catheter of an intravascular ultrasonic system. As shown in FIG. 6, similar to the transducer 100, the silicon ribbon cable 504 is passivated with dielectrics 602 and parylene C 604 such that it can reliably operate in the tissue fluids.

[0032] An embodiment of a fabrication process for an embodiment of a miniature ultrasonic imager array is illustrated in FIGs. 7A-7J using a silicon substrate as an example; however, any material, for example, glass, sapphire, plastic, and ceramic, that is compatible with the thin film process for fabrication of the ultrasonic array is included in the scope of this disclosure. Referring to FIG. 7A, the fabrication process may start with a (100) silicon wafer 700 (where "(100)" refers to the crystal orientation of the silicon wafer) with either n-type or p-type doping. In FIG. 7B, a layer of silicon dioxide 702 may be first thermally grown, followed by the deposition of low stress silicon nitride 704, or other dielectric film, by low pressure chemical vapor deposition (LPCVD), for example.

[0033] In FIG. 7C, on top of the dielectric film, doped polysilicon 706 may be deposited using LPCVD. In addition to doped polysilicon 706, other conducting thin films, such as silicide or doped silicon carbide, can also be chosen to work as the construction material for this layer, the thin film optimally being compatible with subsequent high temperature processes and the final release etch process. The polysilicon film 706 may be annealed to reduce its stress and patterned using a conventional photolithographic step. Using a layer of photoresist as the mask, the polysilicon film 706 can be etched in a dry etcher, forming the counter electrodes, the electrical interconnects, and micro-heaters.

[0034] In FIG. 7D, a layer of sacrificial oxide 708 (for example, phosphosilicate glass) of appropriate thickness (determined by the device specification) may be

deposited and annealed. Photolithographic steps may then be used to create dimple and anchor patterns on the oxide areas and these patterns may be transferred into the oxide layer using dry and/or wet etching. In FIG. 7E, a structural layer of polysilicon 710 is next deposited, doped, and annealed for stress reduction. This polysilicon layer 710 may then be patterned using a conventional photolithographic step and a dry etching, forming membranes and the shielding structures for the electrical interconnects.

[0035] In FIG. 7F, the wafer may then be put in a concentrated HF solution or HF vapor for removing the exposed sacrificial oxide 708. This step will free the membrane so it can move up and down upon application of a voltage on the counter electrode. The release holes 712 in the structural polysilicon may next be sealed using a layer of deposited TEOS oxide. The TEOS film may then be selectively removed using a photolithographic step and a wet etching. This step clears up all TEOS film on the suspended membrane. In FIG. 7G, a layer of dielectric film 714 may next be deposited using LPCVD for protection of the membrane, followed by the deposition of a thin layer of parylene C. A photolithographic step may be used to define the bonding pads.

[0036] In Fig. 7H, a metal evaporation and a lift-off process may then be used to coat metal on these designated areas. A photolithographic step may be used to define the shape of the probe, and all the dielectrics and parylene C 716 outside of the probe areas may be removed using a dry etching. Next a deep dry etching may be performed to dig trenches 718 along the edge of the probe, as shown in FIG. 7I. The wafer may then be flipped and bonded (by gluing or some other bonding mechanism) to a carrier silicon wafer such that the backside of the wafer faces up. The wafer may then be put in a deep silicon etcher for an unmasked etching until the individual device/probe 720

is released from the substrate 722, as shown in FIG. 7J. The carrying wafer and the probes may then be soaked in acetone such that individual probes will be separated from the carrying wafer. A final rinse in IPA may be used to make a probe 720 clean for packaging.

[0037] The silicon ribbon cable shares the fabrication processes with the imager probe except that an additional masking step is needed on the backside of the substrate for defining a different substrate thickness, as shown in FIGs. 8A-K, 9A and 9B. After the completion of the fabrication process for the transducers and electronics, a photolithographic step and a dry etching may be used to create wide trenches on the silicon substrate for device separation. This process at the same time may also create narrow and shallower trenches in the silicon ribbon cable. These shallow trenches may be used to increase the flexibility of the silicon ribbon cable.

[0038] In FIG. 8A, the fabrication process may start with a (100) silicon substrate 800. In FIG. 8B, thermal oxide 802 may be grown, followed by the deposition of low stress silicon 804 nitride in a LPCVD furnace. The thickness of the oxide 802 and the nitride film 804 may be designed such that the overall stress from the composite film is minimized. The dielectric films 802, 804 may be used to insulate the MEMS devices from the substrate. In FIG. 8C, doped polysilicon 806 may be deposited using low pressure chemical vapor deposition (LPCVD). The polysilicon film 806 may then be patterned using a photolithographic process and a dry etching in a reactive ion etcher.

[0039] In FIG. 8D, deposition and patterning of the sacrificial layer 808 may be performed. This layer may be removed in the release step using either a dry or a wet etching. Anchor holes and dimples may be created in this layer using photolithographic steps and dry/wet etching. In FIG. 8E, deposition and patterning of

the structural polysilicon 810 film may be performed. In FIG. 8F, the etching may be released to remove the sacrificial layer. Sacrificial layer 808 may be vacuum sealed to form cavity 812. The release hole may be sealed with TEOS. In FIG. 8G, the passivation layer 814 may be deposited. In FIG. 8H, photolithography and dry etching may be performed to selectively remove the passivation layer and the dielectric films on the substrate 800 to form holes 816.

[0040] In FIG. 8I, after the completion of the fabrication process for the transducers and electronics, a photolithographic step and a dry etching may be used to create wide trenches 818 on the silicon substrate 800 for device separation. This process at the same time also may create narrow and shallower trenches 820 in the silicon for flexibility enhancement of the ribbon cable.

[0041] In FIG. 8J, photolithography and dry etching may be used to create recesses 822 on the backside of the wafer. These recesses 822 may be aligned to map the area for the silicon ribbon cable on the front. In FIG. 8K, An unmasked dry etching may be performed from the backside of the wafer until the trenches 824 on the front side are exposed. This releases individual devices. Because of the unmasking etching which takes away the same amount of silicon everywhere, the recesses 822 created in step 8J on the backside may make the thickness of the silicon ribbon cable thinner than that on the probe area.

[0042] In FIG. 9A, a photolithography and a dry etching may be used to create recesses 902, 904, 906, 908 on the backside of the wafer. These backside recesses may be defined on areas under the silicon ribbon cable. In FIG. 9B, an unmasked dry etching may then be performed from the backside 906 of the wafer until the trenches 902, 908 on the front side are exposed. This may release individual devices. Because of the unmasking etching, which takes away the same amount of silicon everywhere,

the thickness of the silicon ribbon cable may be thinner than that on the probe area because of the recesses created in FIG. 9J on the backside. This process is therefore may be able to create a rigid imager probe and a flexible silicon cable on the same substrate. If more accuracy on the substrate thickness control may be required, boron diffusions and a wet etch in ethylenediamine pyrocatechol or KOH can be used to shape the carrying substrate of the imager probe and the silicon ribbon cable. The hole under the microheater can also be created using a similar approach (boron diffusion and a wet etching or on a recess on the wafer backside like the process used for creating the silicon ribbon cable).

[0043] According to one embodiment, the design of a capacitive ultrasonic transducer as disclosed herein may include the consideration of five specification parameters that are used to determine the performance of the transducer: the resonant frequency of the suspended membrane, the maximum transverse displacement of the membrane, the pull-down voltage, the bias voltage, and the maximum pressure output from the emitter. Both circular and hexagonal membranes may be included in the design. A reason for using a hexagonal transducer may be to gain better area efficiency for a two dimensional array. The five parameters may determine the membrane thickness, diameter of the membrane, and the gap height. For a simple circular membrane, the resonant frequency can be determined by the following analytical equation:

$$\omega_{ns} = \frac{\lambda_{ns}}{R^2} \sqrt{\frac{K}{\mu}}$$

where R is the radius of the membrane of uniform thickness, λ_{ns} is a numerical value for circular membrane, μ is the mass of the plate per unit area, and $K = Et^3/12(1-\nu^2)$ is the bending stiffness (E is the Young's modulus of the membrane material, t is the thickness of the plate, ν is the Poisson's ratio).

[0044] However, as this transducer may be operated in tissue fluids, the influence of damping may be included and a numerical method may be used to determine the resonant frequency. The center deflection of a clamped circular plate under a uniform pressure may be given by:

$$\delta = \frac{1}{64K} P_0 R^4$$

where P_0 is the uniform pressure applied on the membrane. This equation can be used to estimate the pressure of the ultrasonic waves under a prescribed membrane deflection. In this design, the membrane surface displacement required by the specific application may be determined. Three times this displacement plus a safety buffer distance may be used to determine the gap height between the membrane and the counter electrode. The membrane thickness and diameter may then be determined by the surface pressure requirement and the resonant frequency. With all these parameters decided, the pull-down voltage and the bias voltage can be determined. All these parameters and the performance of the transducers can be predicted by numerical simulation before layout design of the devices.

[0045] Any process descriptions should be understood as representing steps in a process, and alternate implementations are included within the scope of the disclosure, in which steps may be executed out of the order described, including substantially concurrently or in reverse order, as would be understood by those reasonably skilled in the art. Further, other systems, methods, features, and advantages of the disclosure will be or become apparent to one with skill in the art upon examination of the drawings and detailed description.

[0046] It should be emphasized that the above-described embodiments, particularly, any “preferred” embodiments, are merely possible examples of implementations,

merely set forth for a clear understanding of the principles of the disclosure. Many variations and modifications may be made to the above-described embodiments without departing substantially in spirit and scope. All such modifications and variations are intended to be included herein within the scope of this disclosure.

CLAIMS

Therefore, at least the following is claimed:

- 1 1. A micromachined ultrasonic system, comprising:
2 a micromachined transducer array; and
3 a micromachined cable that is monolithically integrated with or bonded
4 to the micromachined transducer array.
- 1 2. The system of claim 1, further comprising integrated front-end electronics.
- 1 3. The system of claim 1, further comprising microheaters.
- 1 4. The system of claim 3, wherein the microheaters comprise a temperature
2 sensor and an electrode.
- 1 5. The system of claim 1, wherein the micromachined cable connects the
2 micromachined transducer array to external electronics.
- 1 6. The system of claim 1, wherein the micromachined transducer array and the
2 micromachined cable are fabricated on one substrate.
- 1 7. The system of claim 6, wherein the micromachined cable and micromachined
2 transducer array are fabricated to different thicknesses.

1 8. The system of claim 6, wherein the substrate comprises at least one of silicon,
2 plastic, metal, ceramic, sapphire, or glass.

1 9. The system of claim 6, wherein the substrate is a bio-compatible material.

1 10. A micromachined transducer array, comprising:
2 a substrate;
3 a dielectric film adjacent to the substrate;
4 a thin film electrode adjacent to the dielectric film; and
5 a thin film membrane in contact with the dielectric film and separated
6 from the thin film electrode by a gap.

1 11. The array of claim 10, wherein the electrode and the membrane are fabricated
2 by etching the top and bottom sides of the substrate.

1 12. The array of claim 10, further comprising a passivation layer on the dielectric
2 film and the thin film membrane.

1 13. A method for fabricating a micromachined ultrasonic array, comprising:
2 providing a substrate; and
3 fabricating a micromachined ultrasonic array on the substrate.

1 14. The method of claim 13, wherein the fabricating a micromachined ultrasonic
2 array on the substrate comprises:

3 providing an oxide layer on the substrate;

4 providing a nitride layer on the oxide layer;

5 providing a conducting thin film layer on the nitride layer;

6 patterning the conducting thin film layer;

7 providing a sacrificial layer on the patterned conducting thin film layer;

8 patterning the sacrificial layer;

9 providing a structural conducting thin film layer;

10 patterning the structural conducting thin film layer; and

11 removing the sacrificial layer.

1 15. The method of claim 14, wherein the fabricating a micromachined ultrasonic
2 array on the substrate further comprises creating trenches for device separation.

1 16. The method of claim 13, wherein the substrate comprises a semiconductor
2 material.

1 17. The method of claim 14, further comprising adding a passivation layer.

1 18. The method of claim 14, wherein the substrate is patterned by dry etching on
2 top and bottom sides of the substrate.

1 19. The method of claim 14, wherein the fabricating a micromachined ultrasonic
2 array on the substrate comprises at least one of:
3 boron diffusing;
4 wet etching;
5 dry etching on a silicon-on-insulator substrate; and
6 plastic casting.

1 20. The method of claim 14, further including integrating front-end electronics on
2 the substrate.

1/16

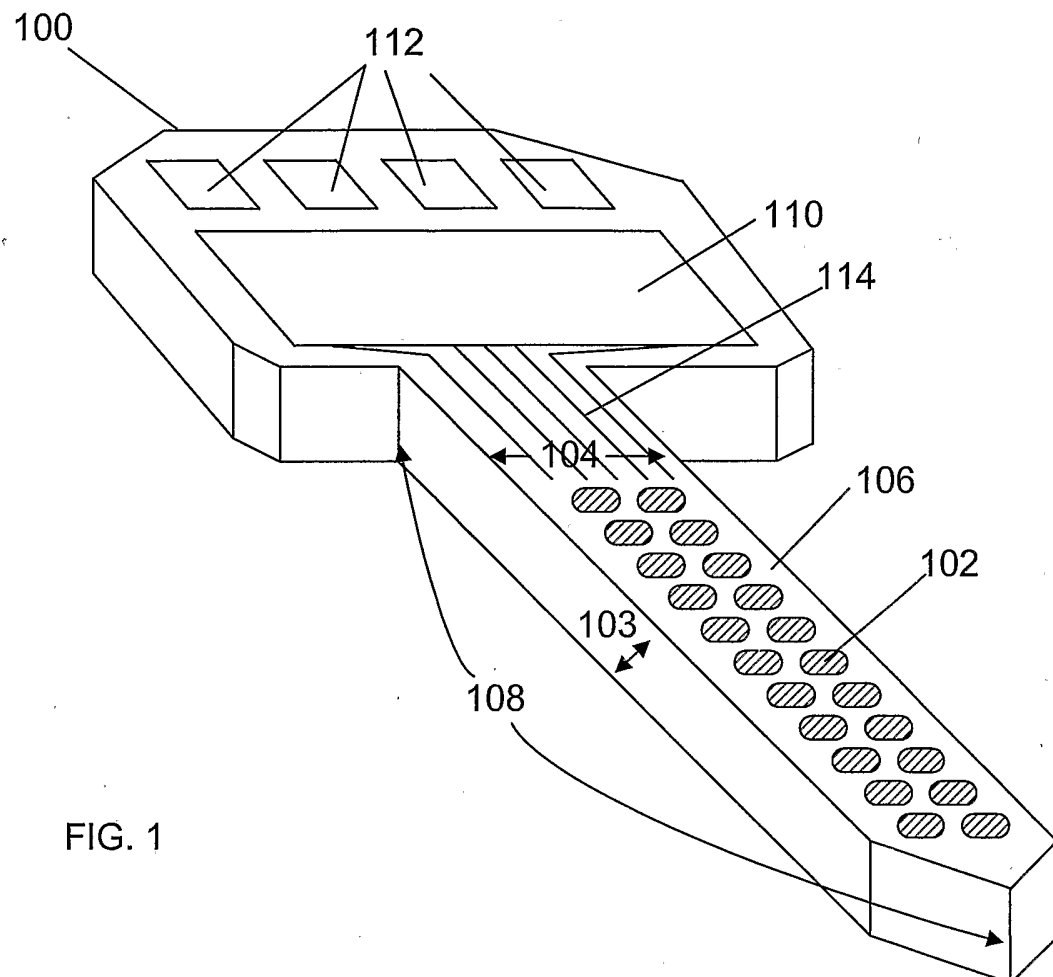
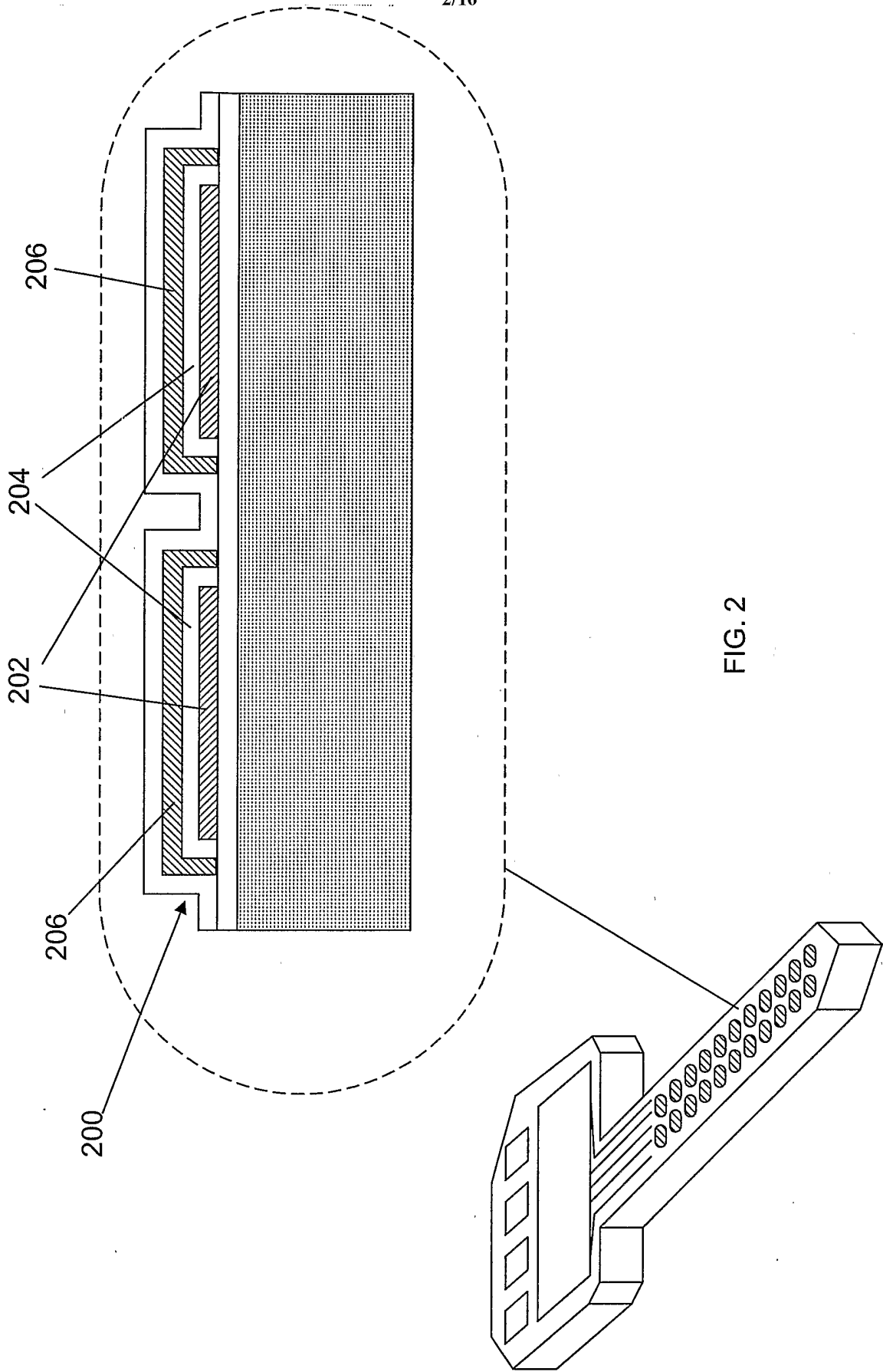
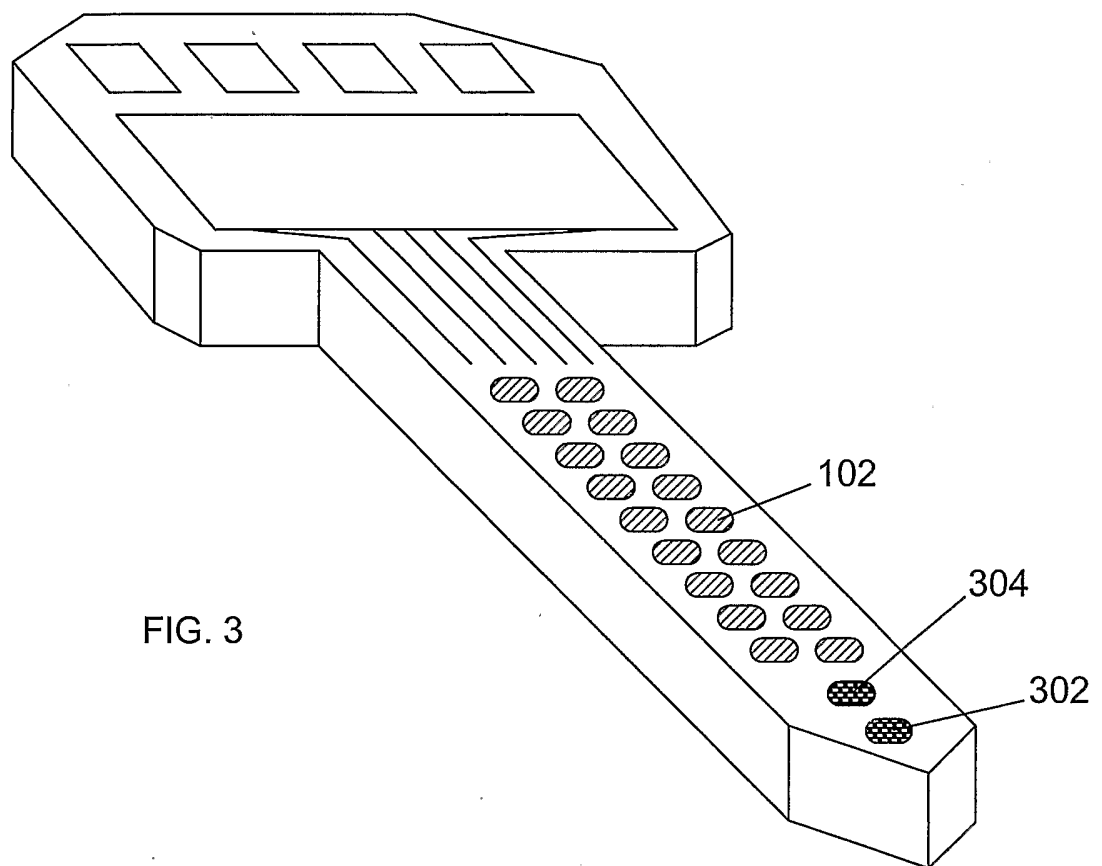
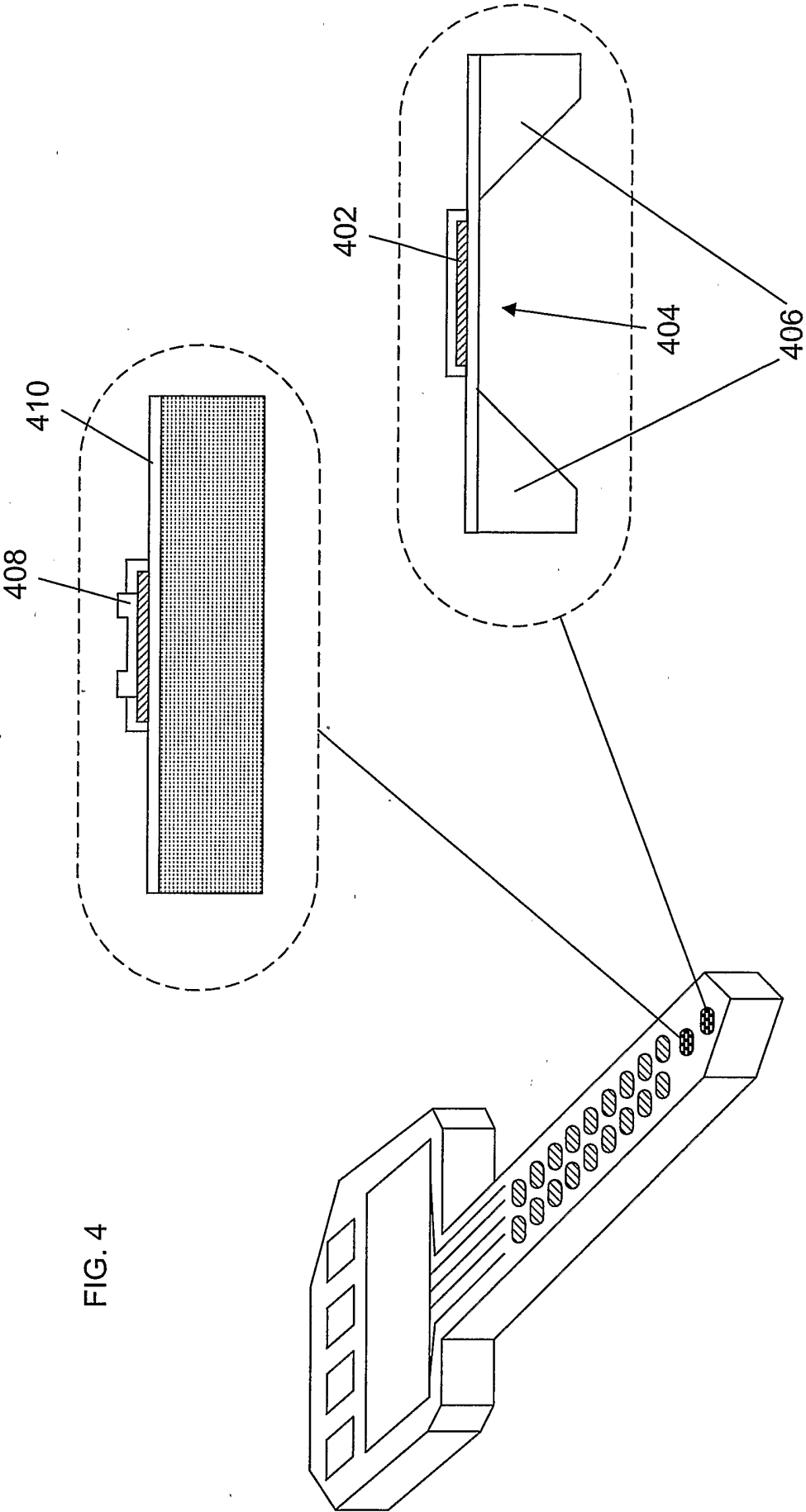
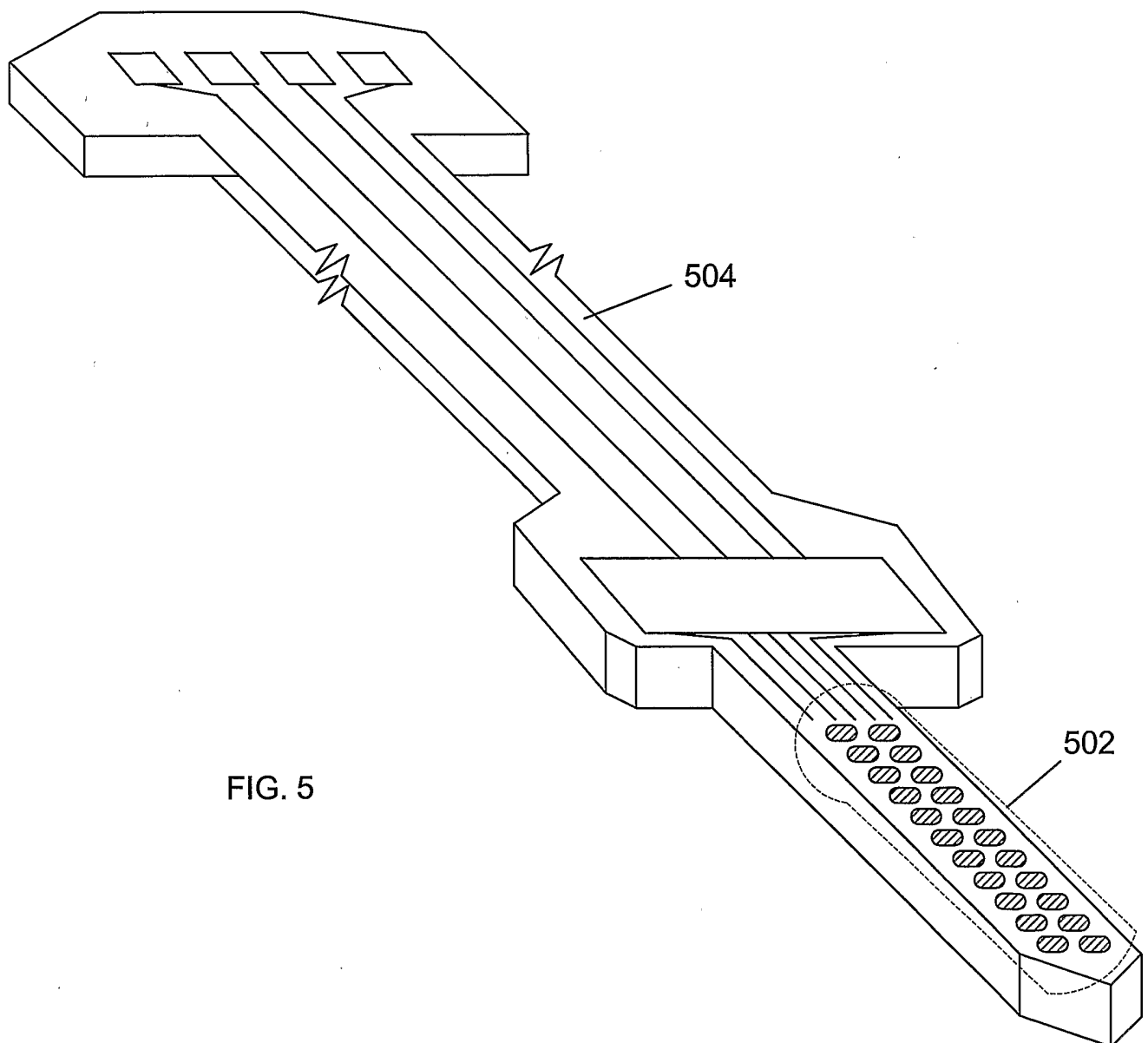


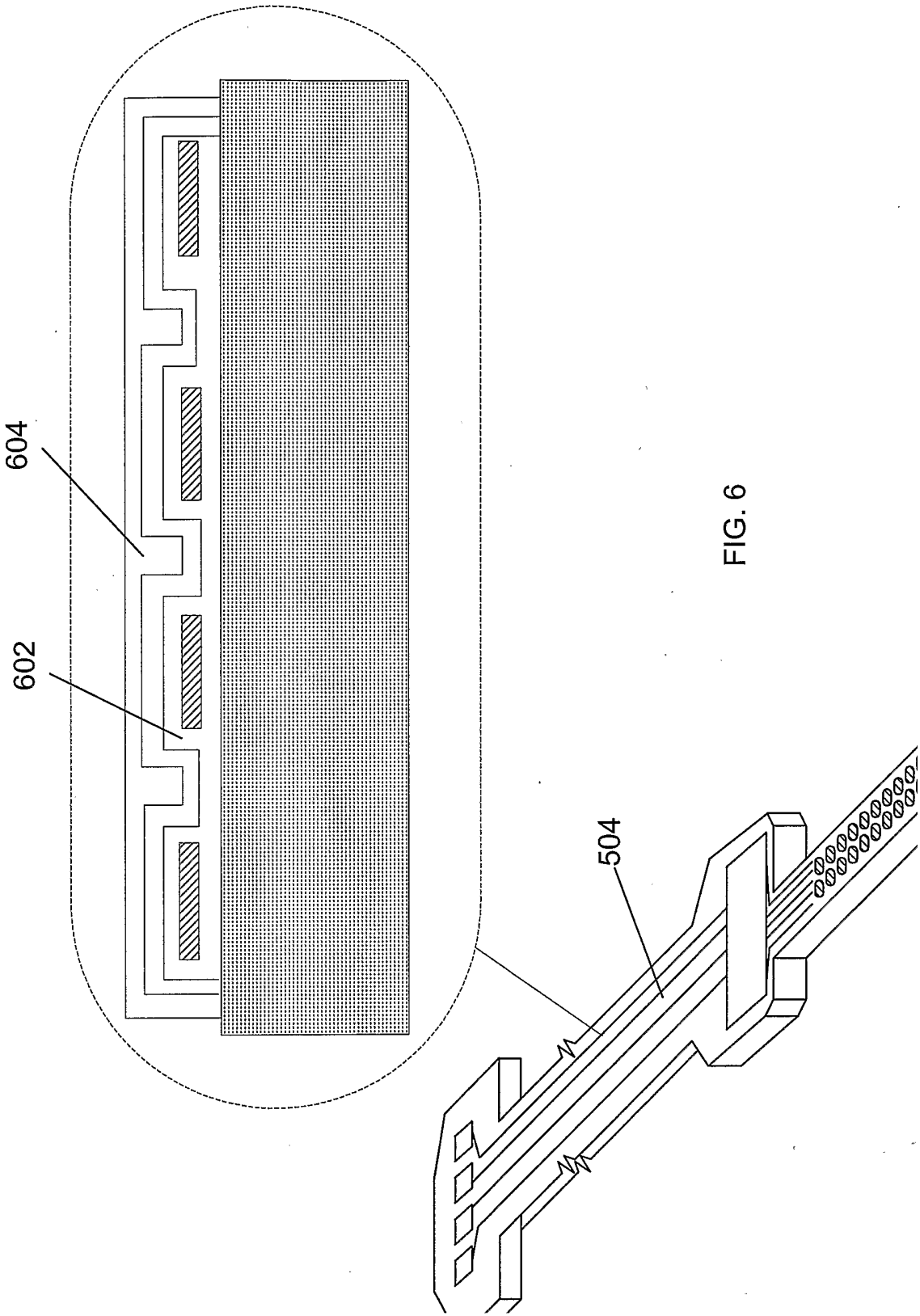
FIG. 1











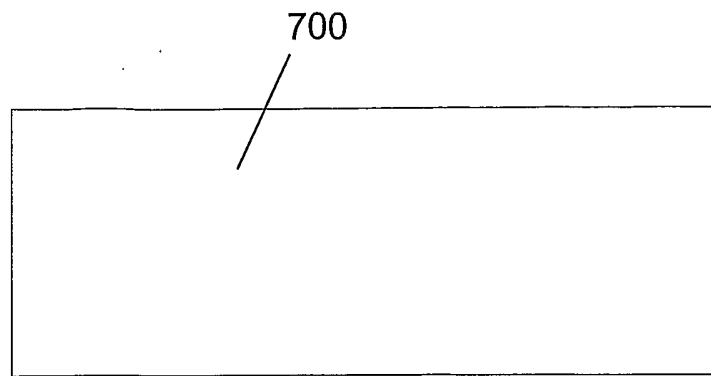


FIG. 7A

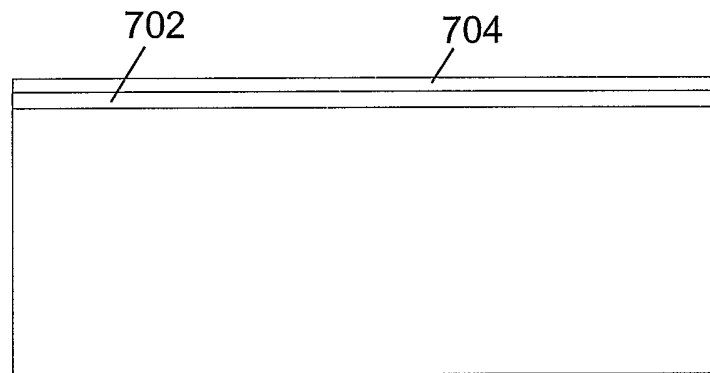


FIG. 7B

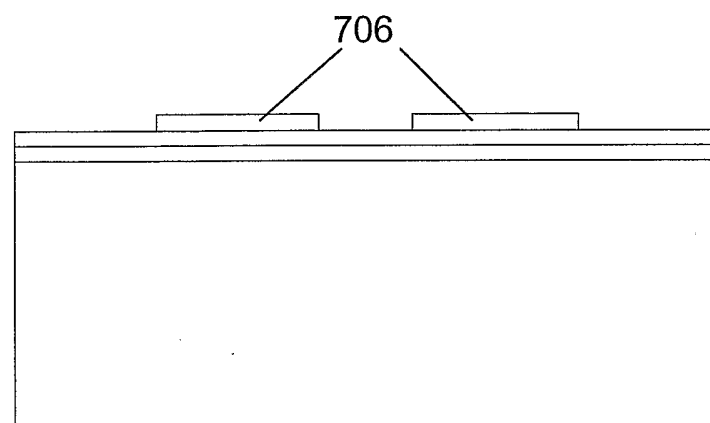


FIG. 7C

FIG. 7D

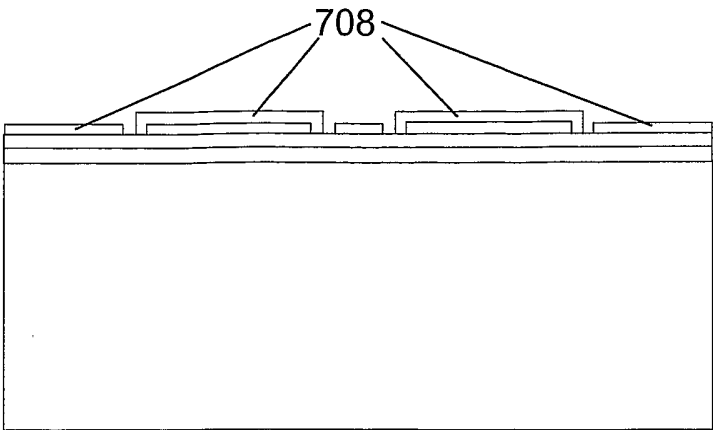


FIG. 7E

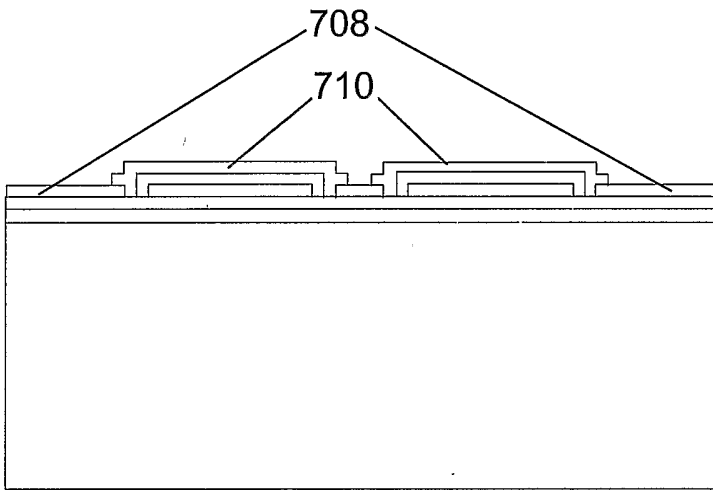
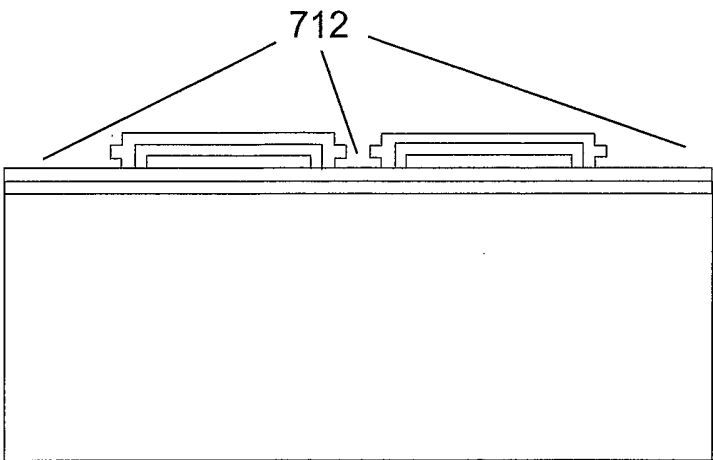


FIG. 7F



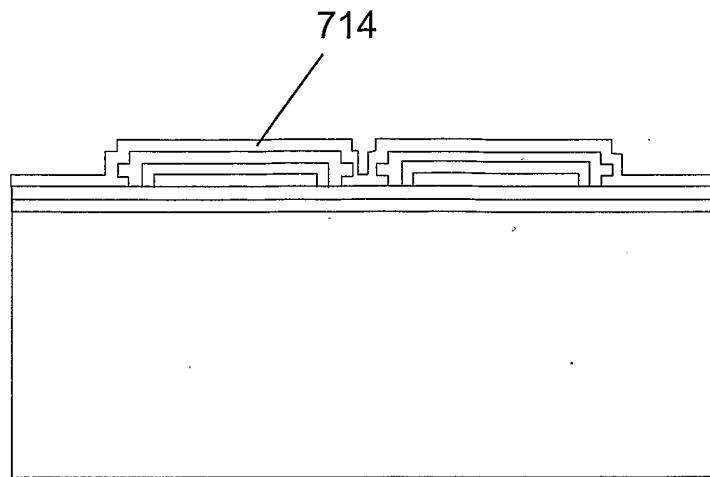


FIG. 7G

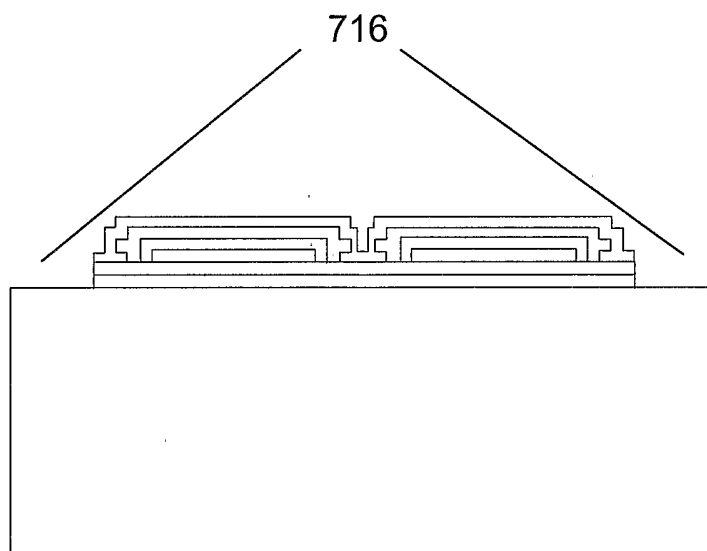


FIG. 7H

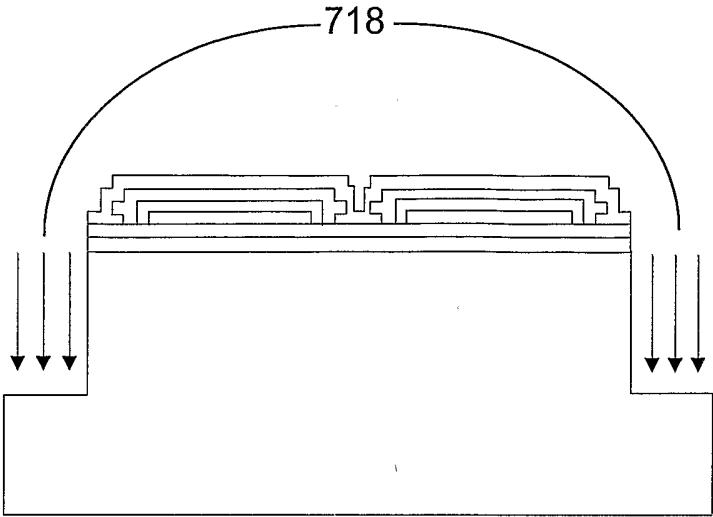


FIG. 7I

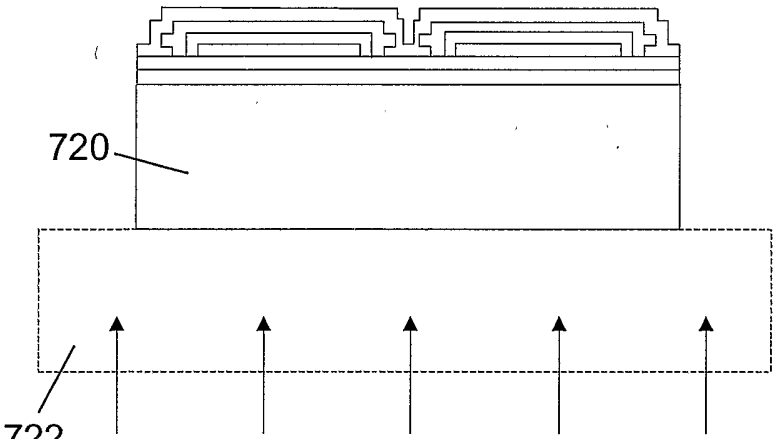
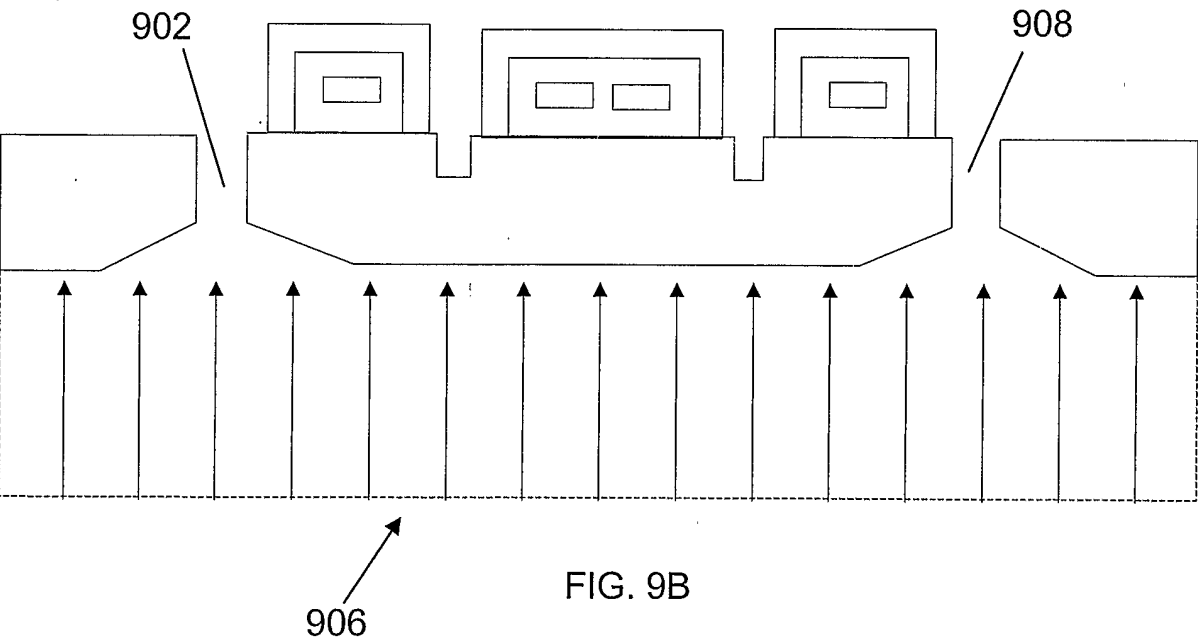
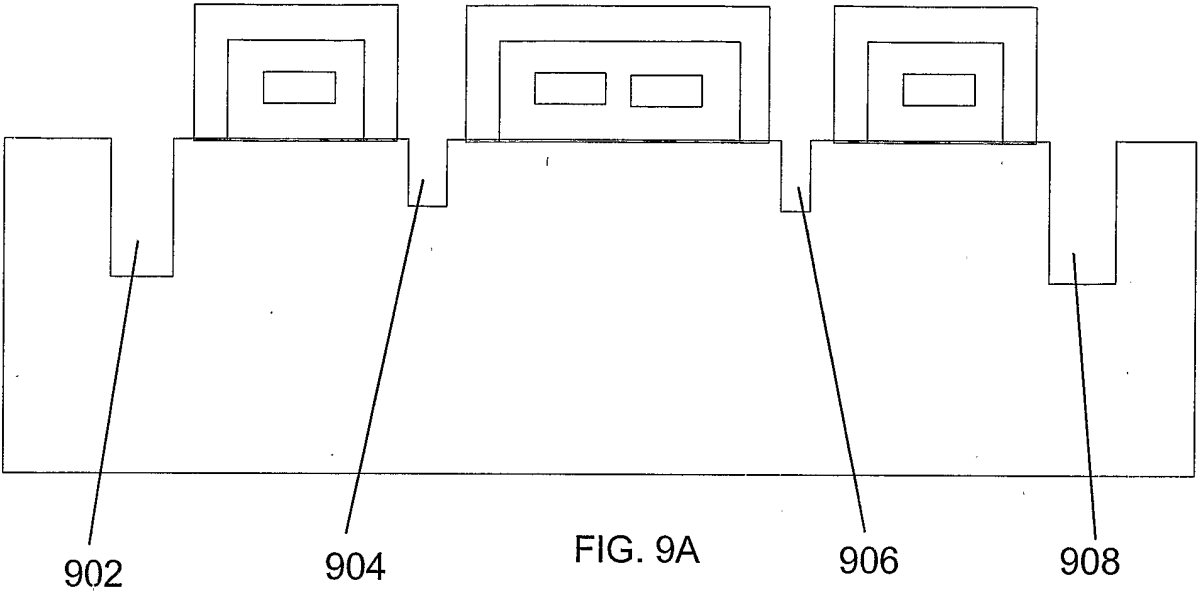


FIG. 7J



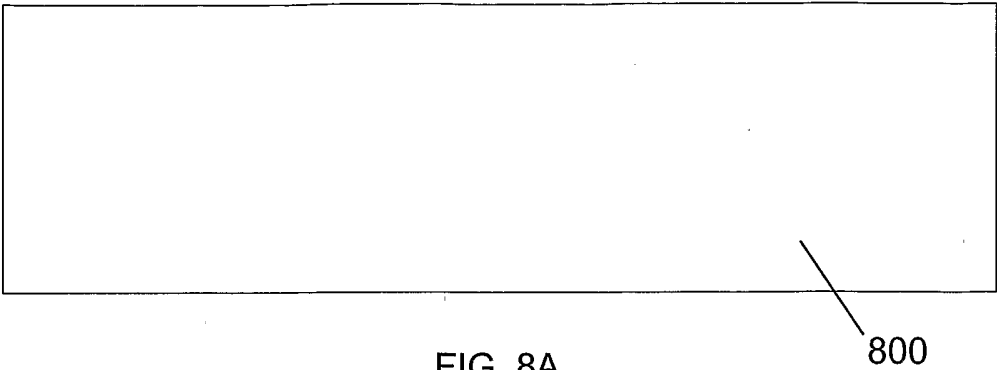


FIG. 8A

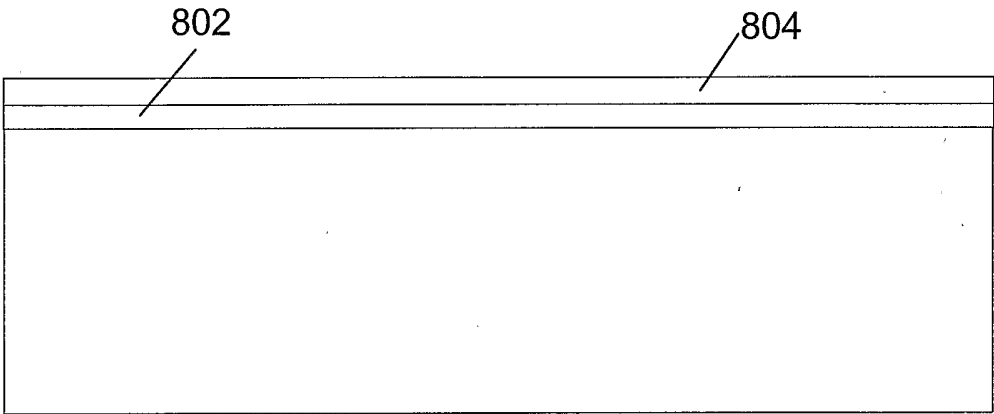


FIG. 8B

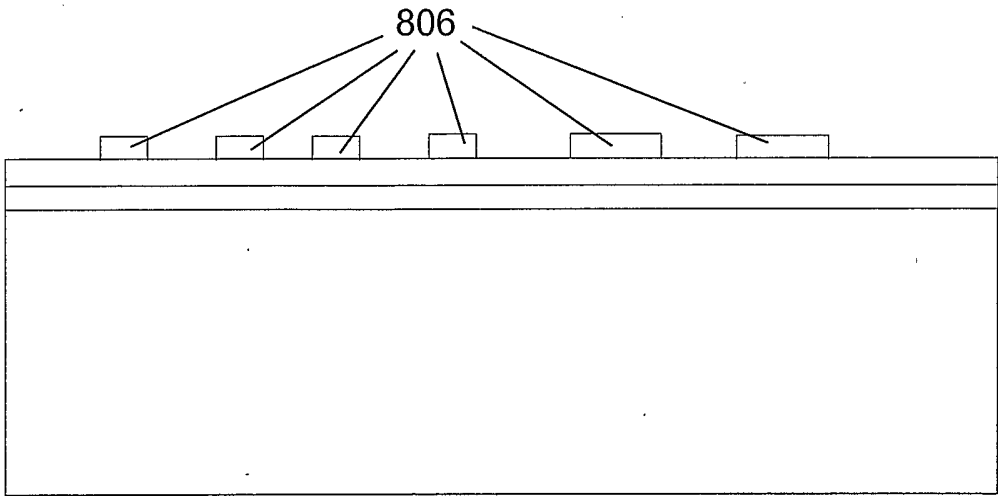


FIG. 8C

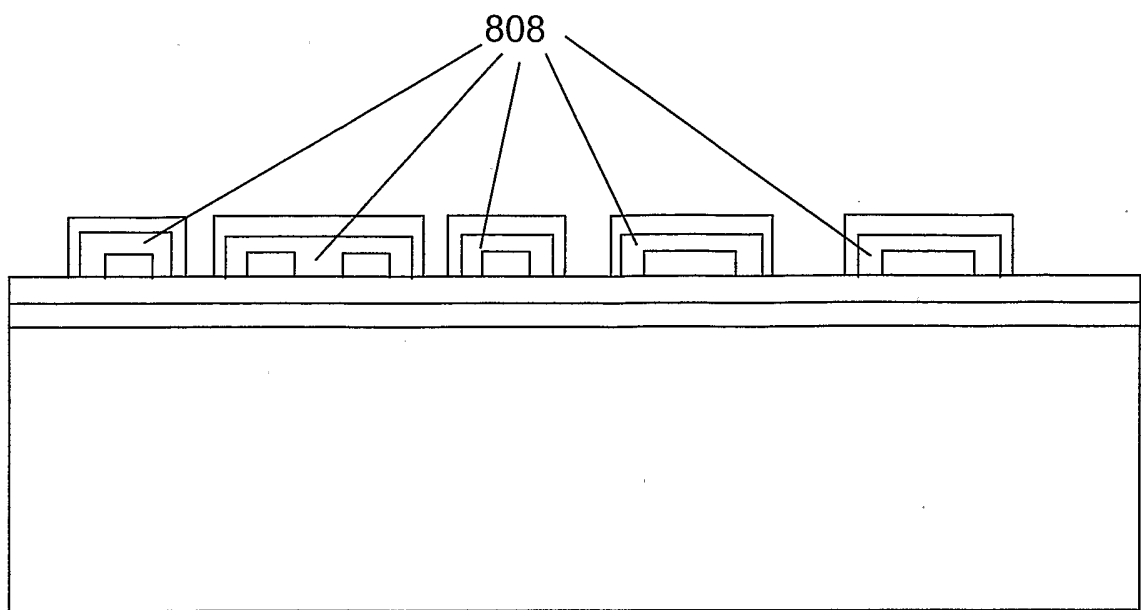


FIG. 8D

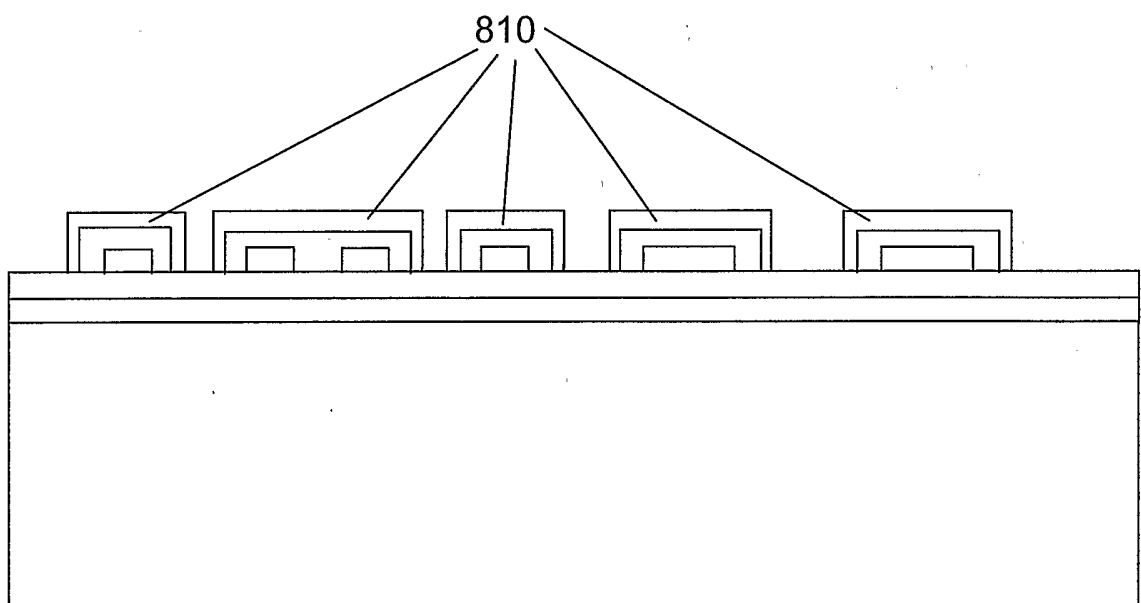


FIG. 8E

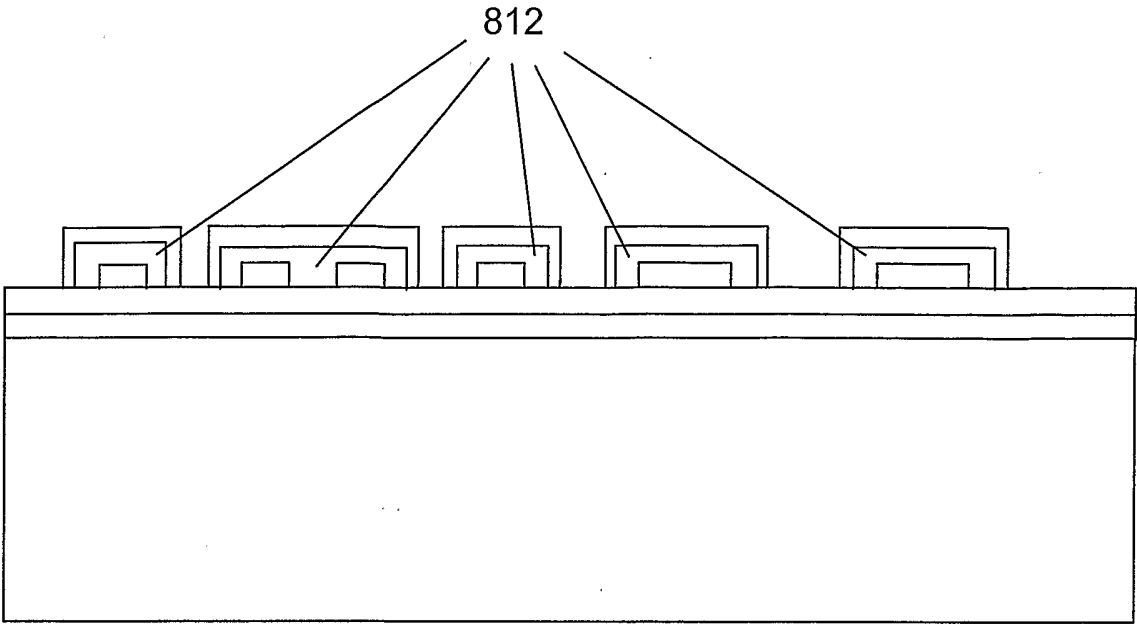


FIG. 8F

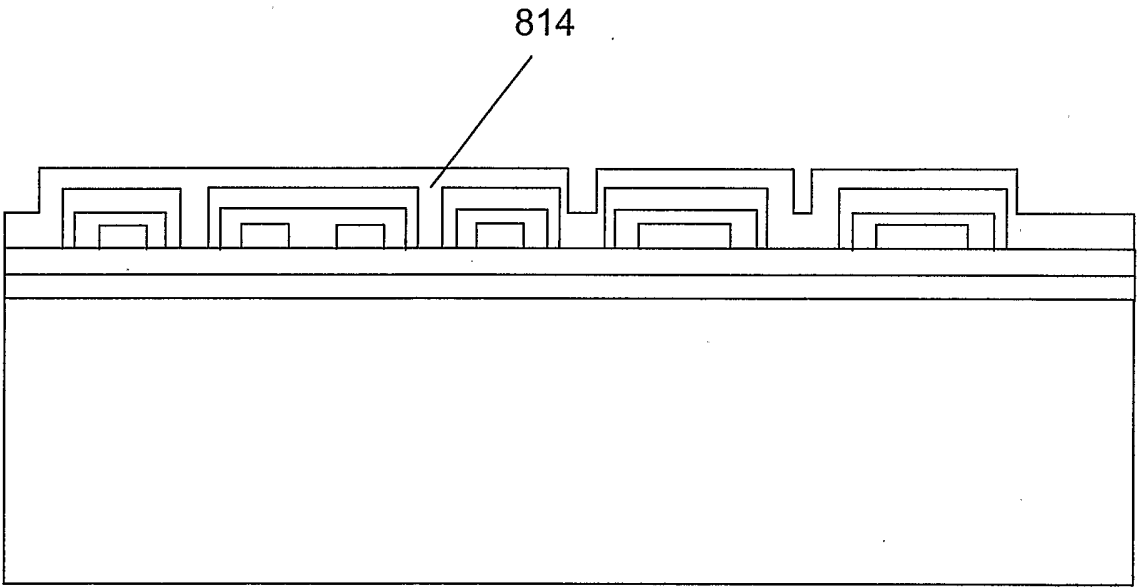


FIG. 8G

