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Acoustic Wave Based MEMS Devices, Development and Applications

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1. Introduction

Acoustic waves based MEMS devices offer a promising technology platform for a wide range of applications due to their high sensitivity and the capability to operate wirelessly. These devices utilize an acoustic wave propagating through or on the surface of a piezoelectric material, as its sensing mechanism. Any variations to the characteristics of the propagation path affect the velocity or amplitude of the wave.

Important application for acoustic wave devices as sensors include torque and tire pressure sensors (Cullen et al., 1980; Cullen et al., 1975; Pohl et al., 1997), gas sensors (Levit et al., 2002; Nakamoto et al., 1996; Staples, 1999; Wohltjen et al., 1979), biosensors for medical applications (Andle et al., 1995; Ballantine et al., 1996; Cavic et al., 1999; Janshoff et al., 2000), and industrial and commercial applications (vapor, humidity, temperature, and mass sensors) (Bowers et al., 1991; Cheeke et al., 1996; Smith, 2001; N. J. Vellekoop et al., 1999; Vetelino et al., 1996; Weld et al., 1999).

This chapter is focused on two important applications of the acoustic-wave based MEMS devices; (1) biosensors and (2) telecommunications. The technological advancement of the micro-electromechanical systems (MEMS) facilitated the development of biosensors and various devices for telecommunications.

There has been increasing interest to develop miniature, portable and low-cost biosensors fabricated using MEMS technologies. For biological applications the acoustic wave device is integrated in a microfluidic system and the sensing area is coated with a biospecific layer. When a bioanalyte interacts with this sensing layer, physical, chemical, and/or biochemical changes are produced. Typically, mass and viscosity changes of the biospecific layer can be detected by analyzing changes in the acoustic wave properties such as velocity, attenuation and resonant frequency of the sensor. An important advantage of the acoustic wave biosensors is simple electronic readout that characterizes these sensors. The measurement of the resonant frequency or time delay can be performed with high degree of precision using conventional electronics.

Currently, a limitation of acoustic wave devices for biological applications is that they require expensive electronic detection systems, such as network analyzers. A final product aimed at the end user market must be small, portable and packaged into a highly integrated cost effective system. For acoustic wave biosensors integrated in a lab-on-chip device, sample

pre-treatment, purification and concentration, as well as a good interface between the user and the integrated sensing system also need to be developed in the future

Historically, acoustic wave devices are widely used in telecommunications industry, primarily in mobile cell phones and base stations. Surface Acoustic Wave (SAW) devices are capable of performing powerful signal processing and have been successfully functioning as filters, resonators and duplexers for the past 60 years. Although SAW devices are technological mature and have served the telecommunication industry for several decades, these devices are typically fabricated on piezoelectric substrates and are packaged as discrete components. The wide flexibility and capabilities of the SAW device to form filters, resonators there has been the motivation to integrate such devices on silicon substrates (Nordin et al., 2007; M. J. Vellekoop et al., 1987; Visser et al., 1989). Standard Complementary Metal Oxide Semiconductor (CMOS) technology with additional MEMS post-processing was used for the fabrication of a CMOS SAW resonator in 0.6 μm AMIs CMOS technology (Nordin et al., 2007). The advantage of using standard CMOS technology for the fabrication of a SAW resonator is that active circuitry can be fabricated adjacent to the CMOS resonator on the same electronic chip.

Telecommunication devices based on acoustic waves have different requirements compared to biosensors. The biosensors operates at frequencies in the range of MHz where acoustic wave devices operating as a filter or resonator are expected to operate at high frequencies (GHz) and have high quality factors and low insertion losses. With the advancement in lithographic techniques, the acoustic wave based devices have the advantage of meeting the stringent requirement of telecommunication industry of having Qs in the 10,000 range and silicon compatibility.

A simple, robust, cheap packaging method is also critical for the commercialization of the acoustic wave devices. The integration of the acoustic wave based MEMS biosensor in the microfluidic system is a complex matter. The integration technique is influenced by the sensor fabrication process and the type of the biological applications. In some applications the sensor could be embedded in a microfluidic reservoir. In the case when the biological application requires different biological solutions to be introduced on the sensor sensitive area the biosensor could be embedded in a microfluidic channel. The packaging of the acoustic wave devices used for telecommunication is less complicated since these devices are embedded in the package and do not need to be in contact with liquid.

2. Acoustic wave MEMS devices as biosensors

There has been increasing interest to develop miniature, portable and low-cost biosensors fabricated using MEMS technologies. MEMS technology has been adopted from the integrated circuit (IC) industry and applied to the miniaturization of a large range of systems including acoustic wave based devices. Recent technological advancement of MEMS processes allows the fabrication of thin piezoelectric films and the integration of acoustic wave based devices, and electronics on a common silicon substrate. The acoustic wave MEMS biosensors presented in this chapter could be categorized in two main groups; resonators and delay lines.

For biological applications the Acoustic Wave Based MEMS devices are integrated in a microfluidic system and the sensing area is coated with a biospecific layer. When a bioanalyte interacts with this sensing layer, physical, chemical, and/or biochemical changes are produced. Typically, mass and viscosity changes of the biospecific layer can be detected

by analyzing changes in the acoustic wave properties such as velocity, attenuation and resonant frequency of the sensor. An important advantage of the acoustic wave biosensors is simple electronic readout that characterizes these sensors. The measurement of the resonant frequency or time delay can be performed with high degree of precision using conventional electronics.

The Sauerbrey equation correlates the changes of the resonant frequency of an acoustic wave resonator with the mass deposited on it. The acoustic wave propagating on a piezoelectric substrate is generated and received using IDTs. In the case of a biosensor resonator, the cell to be analyzed or the antibody layer for protein marker detection are added on the IDTs. This will cause a shift of the resonant frequency due to the increasing of mass, where f_i and f_o are the resonant frequencies before and after loading the sensor.

The Sauerbrey equation is defined as;

$$\Delta f = -\frac{2f_o^2 \Delta m}{A\sqrt{\rho_q \mu_q}} = -2.26 \cdot 10^6 f_o^2 \frac{\Delta m}{A} \quad (1)$$

where

$$f_o - f_i = \Delta f \quad (2)$$

From (1) the change Δf of the resonant frequency of the piezoelectric crystal is directly proportional to the mass loaded on the acoustic wave resonator, where Δm is expressed in g and Δf and f_o in Hz (Skládal, 2003).

Generally, the acoustic wave MEMS resonators employed for biosensing applications are FBAR and acoustic wave based delay lines. For FBAR type biosensor the excitation electrodes are fabricated at both sides of the piezoelectric substrate and the acoustic waves propagate through the volume of the substrate. The detection mechanisms occur at the opposite surfaces of the piezoelectric substrate.

The acoustic wave based delay lines reported in the literature as MEMS biosensors are surface acoustic wave (SAW) delay lines that consists of two sets of interdigitated transducers (IDT)s fabricated on the same side of a thin layer of piezoelectric material. The acoustic wave is produced by one set of IDTs and the second set of IDTs is used to detect the acoustic wave. In the case of a biosensor, the surface between these two sets of IDTs is covered with a biological layer sensitive to the analyte to be detected, as illustrated in Fig. 1.

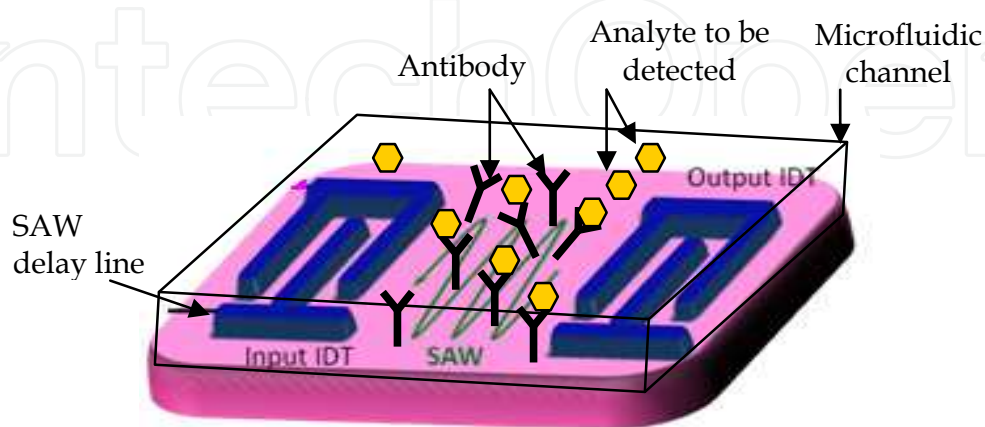


Fig. 1. SAW delay line biosensor integrated in a microfluidic channel. The surface between the IDTs is coated with antibodies sensitive to the analyte to be detected. The analyte

molecules binding to the immobilized antibodies on the sensor surface influence the velocity of the SAW and hence the output signal generated by the driving electronics.

The absorption of the analyte on the sensitive layer will produce a time delay in the acoustic wave propagation. The main disadvantage of the acoustic wave based devices when used as biosensors is the degradation of performance due to liquid damping. In liquid the quality factor Q drops (usually more than 90% reduction) and negatively affects the device sensitivity. Since most of the biological applications are performed in liquid only few types of acoustic wave devices could be integrated in microfluidic channels, without significant degradation of the sensor performance.

2.1 Film Bulk Acoustic wave Resonators (FBAR)

In recent years the thin film fabrication technology has made substantial progress particularly in view of high frequency resonators. In the case of MEMS-based FBAR resonators the expensive single crystalline substrates used for quartz crystal microbalance (QCM) resonators, could be replaced with a large range of thin piezoelectric films. FBAR resonators that are fabricated from a thin piezoelectric substrate and the excitation electrodes are fabricated at both sides of the piezoelectric substrate. The FBAR resonators could be integrated in a microfluidic system and successfully used for biosensing applications because of low damping of the acoustic wave in the liquid (Gabl et al., 2003; Weber et al., 2006; Wingqvist et al., 2005).

A MEMS FBAR biosensor with aluminum nitride (AlN) as piezoelectric film is illustrated in Fig. 2 (Wingqvist et al., 2005). The thickness of the AlN film is $2\mu\text{m}$. Bottom and top Al electrodes were patterned with standard lithography and etching processes. The overlap of the top and bottom electrode defines the active area where the acoustic wave is generated. In order to fabricate a shear acoustic wave FBAR, the AlN thin films were grown with the crystallographic axis inclined with an angle of 30° relative to the surface normal. The shear mode is preferred for liquid application instead of the longitudinal mode because of low-loss operation in liquid and small reduction of the quality factor Q . The silicon wafer was etched from the back side to fabricate a free standing membrane used to isolate the resonator acoustically from the substrate and define a cavity. This cavity was further connected to a microfluidic transport system for analyte delivery to the bottom electrode of the resonator. For this type of FBAR, the bottom electrode is the sensing electrode. An Au layer was thermally evaporated onto the bottom Al electrode to create a biochemically suitable surface for subsequent tests. The sensor was tested with different concentration of albumin in solution and the detection limit was 0.3 ng/cm^2 .

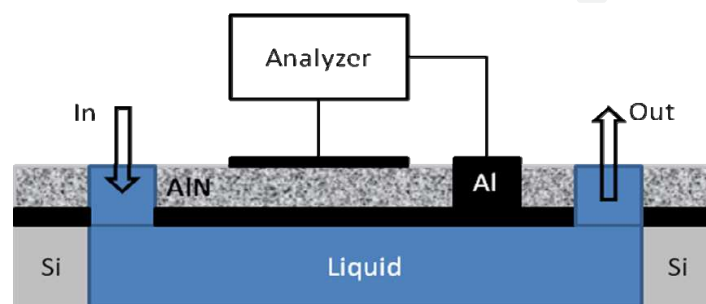


Fig. 2. Schematic of a shear wave FBAR with a microfluidic transport system.

Another FBAR resonator used for biological applications, illustrated in Fig. 3, is fabricated from a thin film of ZnO (Zhang et al., 2009). This FBAR resonator was fabricated from a <100> silicon wafer. A thin silicon dioxide layer was thermally grown on the wafer, followed by the low pressure chemical vapor deposition (LPCVD) of a thin SiN layer. An Al film was evaporated and patterned on the SiN layer as the bottom electrode and determines the effective area of the FBAR. The ZnO piezoelectric layer with the thicknesses from 0.55 μm to 4.2 μm was sputtered on the Al bottom electrode. Layers of Cr/Au representing the top electrode were sputtered and patterned by lift-off over the ZnO layer. Au was chosen as the top electrode due to its excellent conductivity and good affinity for biomolecular binding. The microfluidic channel was fabricated from a 3 μm thick parylene film deposited on the top electrode. The silicon wafer was backside etched using deep reactive ion etch (DRIE) to release the SiN membrane. The SiO₂ layer underneath the SiN layer was removed by wet etching, see Fig. 3a. The top Au electrode of the FBAR represents the sensing electrode and in the case of biological applications will be coated with a thin layer of biological material. This sensor is demonstrated a quality factor Q in liquid of 120. Q is improved by integrating a microfluidic channel on FBAR, which confines the liquid to a height comparable to the acoustic wavelength. However, this FBAR biosensor is sensitive to temperature variations. Increasing the temperature degrades Q , resulting in degradation of resolution.

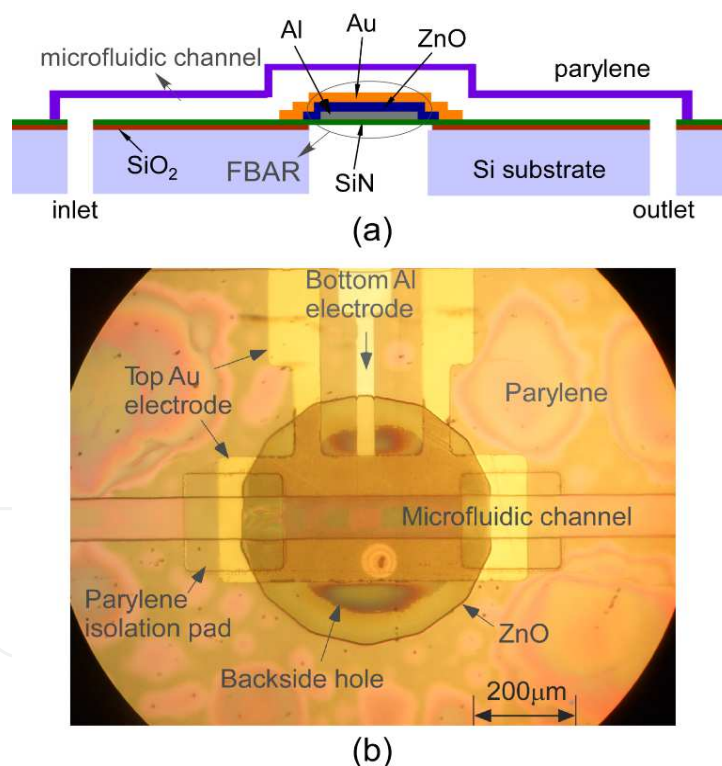


Fig. 3. (a) Schematic of the FBAR sensor integrated with a microfluidic channel and (b) Top view of the fabricated FBAR sensor; a microfluidic channel run across the FBAR sensor (Zhang et al., 2009). (© [2009] IEEE.) Used with permission.

Typical MEMS FBAR's have high quality factor Q in the air and vacuum which range from a few hundreds to a few thousands. High Q s are possible since the acoustic wave generated in the FBAR is well confined by the very large acoustic impedance mismatch between the solid materials and air. However, for biological applications the FBAR resonators are immersed in

liquid and the solid-liquid interface becomes leaky for acoustic waves since the impedance mismatch is small, resulting in the degradation of Q of FBAR. To improve Q of FBAR in liquid, it is recommended that the microfluidic channels on top of the FBAR have the heights comparable to the acoustic wavelength in FBAR. This minimizes the dissipation in the liquid and hence improves the resonator's Q .

An interesting design of an FBAR without significant reduction of Q factor in liquid environments is illustrated in Fig. 4 (Pottigari et al., 2009). This device employs a thin vacuum gap between the FBAR and the sensing interface to prevent acoustic energy loss in liquid. This vacuum gap acts as an acoustic energy loss isolation layer of the FBAR in the aqueous viscoelastic media and reduces the direct contact area at the interface between FBAR and liquid. This vacuum separation is achieved by using micro-posts between the top electrode and the sensing diaphragm, which is in contact with the liquid. When the liquid is loaded on the sensing diaphragm, the mass is directly transferred onto the FBAR through the microposts. The vacuum gap protects the FBAR surface from the liquid, prevents the acoustic energy loss from the liquid, and contributes to maintenance of a high Q factor in the liquid without losing mass sensitivity.

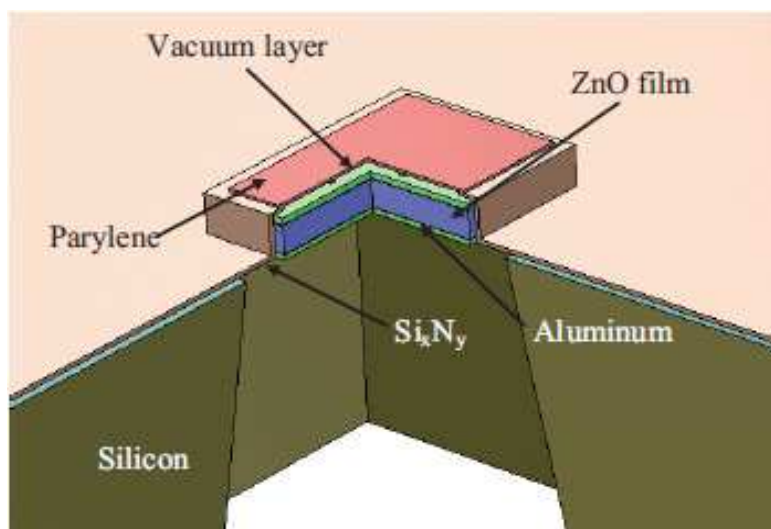


Fig. 4. Cross-section view of the V-FBAR device (Pottigari et al., 2009).
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This vacuum (V)-FBAR uses a $0.7\ \mu\text{m}$ thick ZnO film as piezoelectric material. The ZnO piezoelectric film is deposited by a radio-frequency (RF) sputtering system. A $1.6\ \mu\text{m}$ thick parylene layer forms the top sensing diaphragm. Fig. 5 shows the V-FBAR integrated with a parylene sensing diaphragm that is supported by several parylene microposts. The height of the vacuum space is $2\ \mu\text{m}$ and the active sensing area is $200\ \mu\text{m} \times 200\ \mu\text{m}$ (Pottigari et al., 2009). The sensing diaphragm fabricated over a vacuum gap and the microposts are enhancing the sensitivity of the FBAR by separating the liquid damping effect from the operating frequency of the device. This device was conceived for biological applications.

Another recently developed MEMS based FBAR-type biological sensor is illustrated in Fig. 6 (Gabl et al., 2003). This FBAR is formed by two electrodes, fabricated at both sides of a thin a piezoelectric layer and integrated above a silicon substrate. The active vibrating region of the resonator is coated with a receptor layer which is sensitive to the biological analyte to be detected. The attachment of the analyte molecules leads to an increase of the resonator mass load and a decrease of the resonant frequency which can be electrically determined.

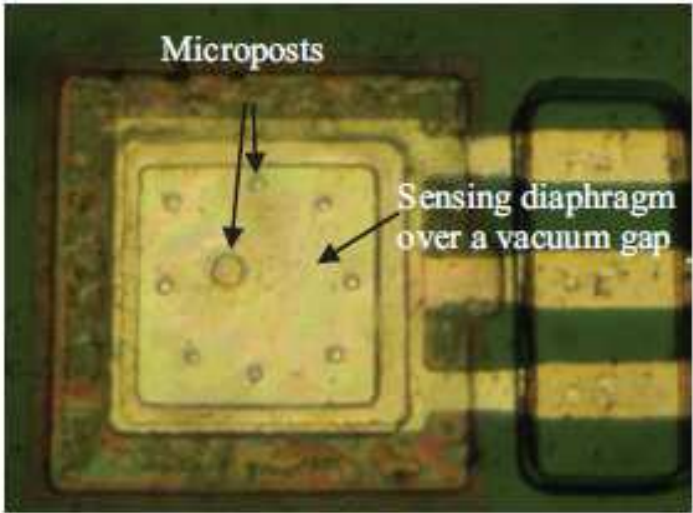


Fig. 5. Photo of fabricated V-FBAR (Pottigari et al., 2009).
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The FBAR sensors have been fabricated on silicon substrates employing reactive magnetron sputtering of ZnO (Fig. 7). The 100 nm Au top-electrode provides a low electrical series resistance as well as a common chemical base for the binding of bioreceptor molecules. The quarter wavelength thick bottom electrode acts acoustically as an efficient reflection layer and ensures a high mass sensitivity as well as a low ohmic series resistance. A 3-fold ZnO/Pt mirror is fabricated below the FBAR sensor in order to isolate the sensor from the silicon substrate, see Figs. 6 and 7.

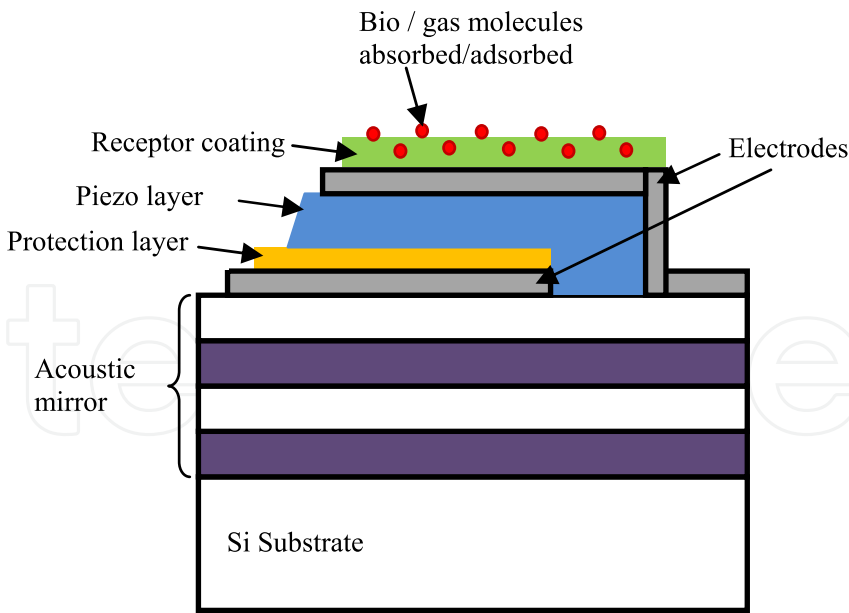


Fig. 6. Illustration of FBAR type biological sensor.

To verify the use of this type of FBAR in biosensing, a receptor assay of biotin-labeled DNA oligos coupled to the gold surface and streptavidin as the target molecule, were used. This FBAR biosensor is characterized by a sensitivity three orders of magnitude larger than for typical QCM.

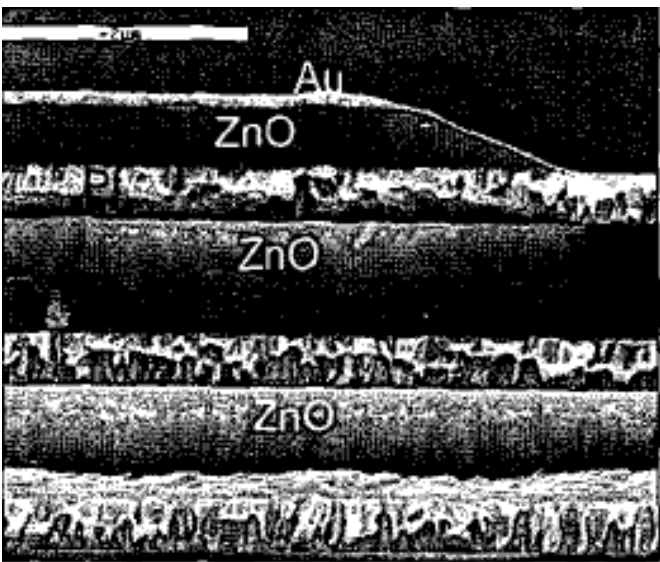


Fig. 7. SEM image of the acoustic mirror and ZnO thin films on a Si substrate. (Gabl et al., 2003). (© [2003] IEEE). Used with permission.

A nano-FBAR biosensor that uses nanomaterials to increase its sensitive area and it is based on $Mg_xZn_{1-x}O$ as piezoelectric material was reported in the literature (Chen et al., 2009). This device is built on Si substrates with an acoustic mirror consisting of alternating quarter-wavelength silicon dioxide (SiO_2) and tungsten (W) layers to isolate the FBAR from the Si substrate, see Figs. 8 and 9. High-quality ZnO and $Mg_xZn_{1-x}O$ thin films are achieved using RF sputtering technique. Tuning of the device’s operating frequency was realized by varying the Mg composition in the piezoelectric $Mg_xZn_{1-x}O$ layer. The ZnO nanostructures were grown on Au electrode situated on the TFBAR’s top surface using metalorganic

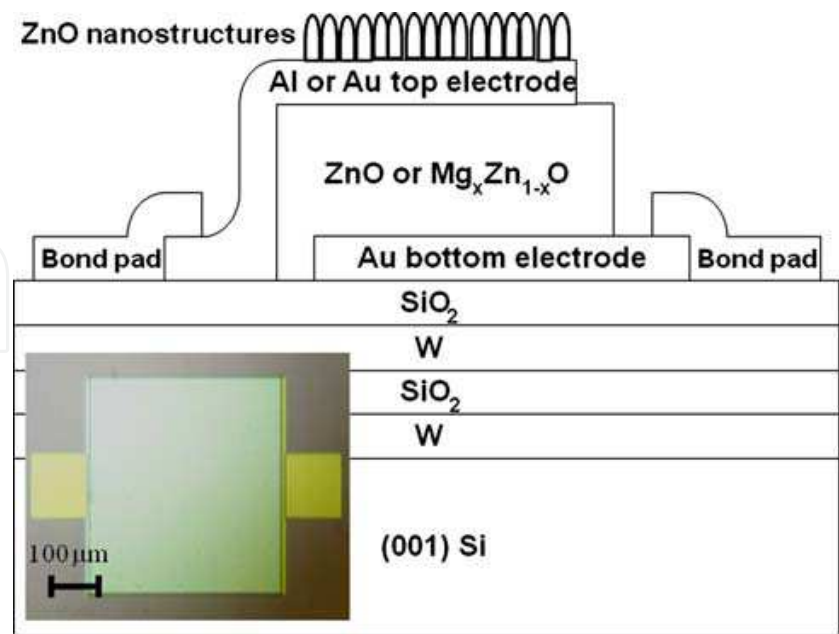


Fig. 8. Schematic diagram of the $Mg_xZn_{1-x}O$ FBAR structure; the inset shows a micrograph of a FBAR device. For the FBAR nanosensor, Au top electrode is used to facilitate MOCVD growth of ZnO nanostructures (Chen et al., 2009). (© [2009] IEEE). Used with permission.

chemical vapor deposition (MOCVD). These nanostructures offer a very large sensing area, faster response, and higher sensitivities over the planar sensor configuration. Because of the large sensitive area, the mass sensitivity of this biosensor is higher than $103 \text{ Hz cm}^2/\text{ng}$. In order to employ this nano-FBAR for biosensing, the nanostructured ZnO surface was functionalized to selectively immobilize DNA. The device sensitivity was 16.25 ng of hybridized DNA and linker molecules combined.

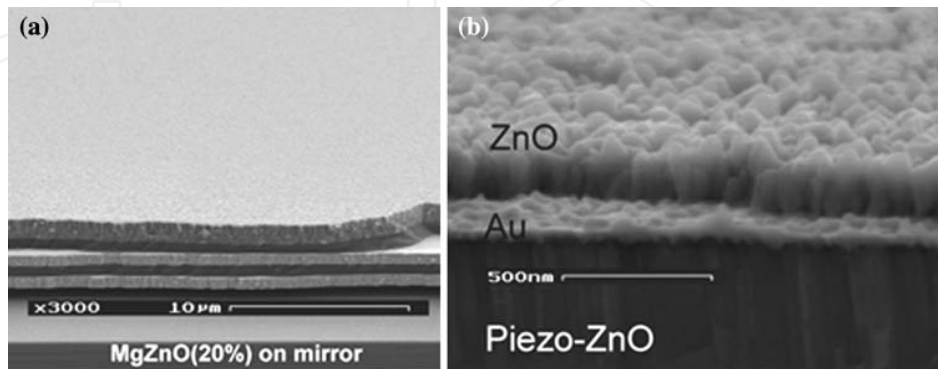


Fig. 9. Cross-sectional SEM images of (a) $\text{Mg}_{0.2}\text{Zn}_{0.8}\text{O}$ film deposited on the mirror/Si structure and (b) ZnO nanostructures deposited on Au electrodes (Chen et al., 2009). (©[2009] IEEE). Used with permission.

The FBAR can be modeled into an equivalent circuit using a constant “clamped” capacitance C_0 connected in parallel with an acoustic (motional) arm that consists of motional capacitance C_m , motional inductance L_m , and motional resistance R_m , as illustrated in Fig. 10. The formulas for each component are given below:

$$C_0 = \epsilon_r \epsilon_0 \frac{A}{d}, \quad (6)$$

where A is the area of overlap of the two electrodes, ϵ_r is the relative static permittivity of the material between the electrodes, ϵ_0 is the electric constant and d is the separation between the electrodes.

$$C_m = \left[\left(\frac{f_p}{f_s} \right)^2 - 1 \right] \quad (7)$$

where f_p is the parallel resonant frequency and f_s is the series resonant frequency.

$$L_m = \frac{1}{(2\pi f_s)^2 C_m} \quad (8)$$

$$R_m = \frac{1}{2\pi f_s C_m Q} \quad (9)$$

where Q is the quality factor of the resonator.

The FBAR biosensor is operated based on the dependency of the resonator frequency on its mass. The biomolecules attached on the sensing electrode increase the sensor mass. The sensitivity S (frequency shift per mass attachment) is described as:

$$S = \frac{f_0}{M} \tag{10}$$

where f_0 is the operating frequency and M is the resonator mass [11].

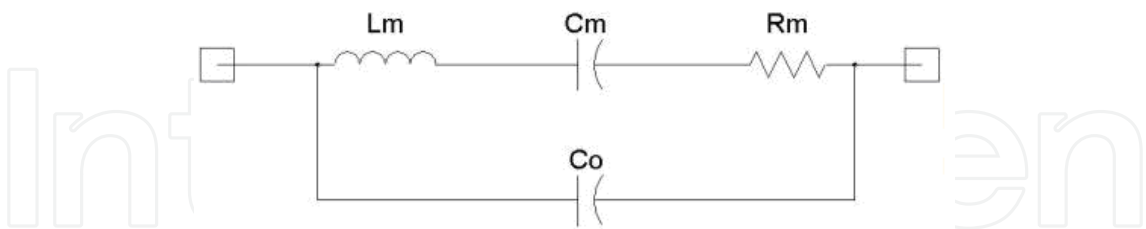


Fig. 10. FBAR equivalent circuit, where L_m is the motional mass, C_m is the motional capacitance, R_m is the motional resistance and C_0 is the static capacitance.

2.2 SAW delay lines as biosensors

Surface acoustic wave (SAW) delay lines were also studied for biosensing and could be integrated in microfluidic systems. A SAW delay line consists of two IDTs that are electrode pairs fabricated on the same side of a thin piezoelectric layer via photolithography. A sinusoidal voltage applied to the input IDTs translates into oscillating mechanical strain that forms a SAW that propagates along the surface of the piezoelectric thin film. The SAW is then

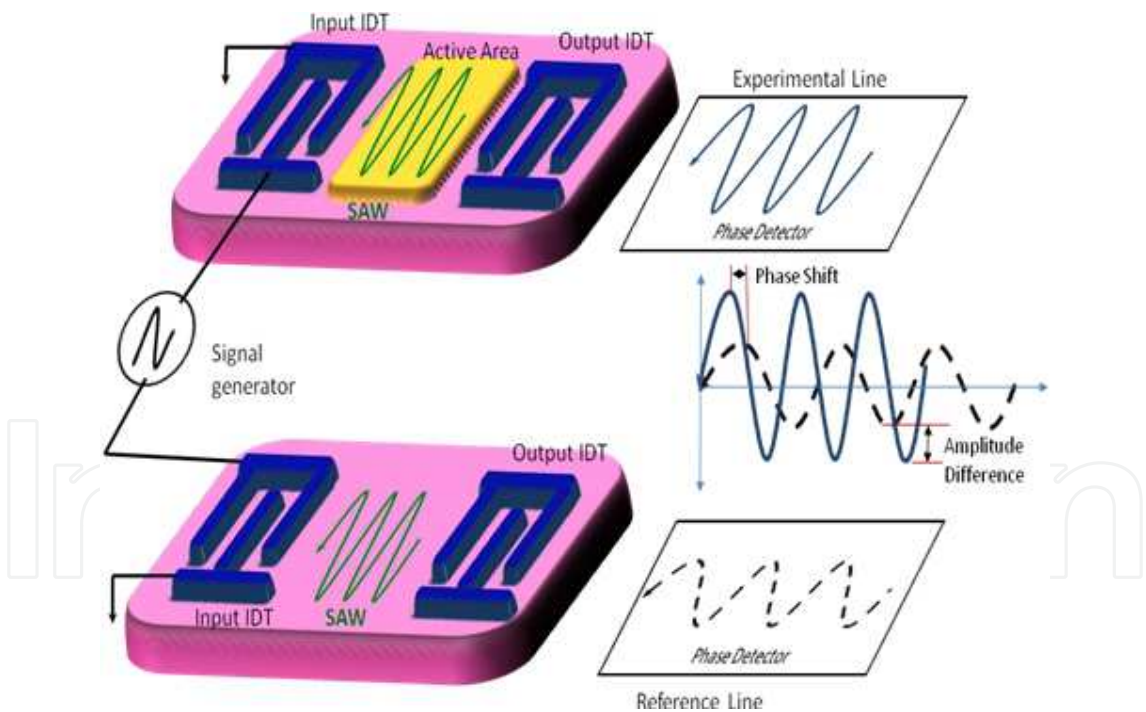


Fig. 11. A SAW delay line consists in input IDT and output IDT fabricated on a piezoelectric substrate. Two delay lines operate in parallel, with one line acting as a reference line and the other acting as an experimental line. A sinusoidal voltage is applied to the input IDT, which develops an alternating electric field that is translated into a mechanical SAW by the piezoelectric effect. The velocity of the SAW is affected by the mass loading, the liquid viscosity and the temperature of the substrate surface. Any difference in velocity between the two delay lines will be reflected as a phase shift and amplitude difference.

converted back into a sinusoidal voltage of different frequency (phase) and amplitude at the output IDTs. These differences are related to changes in the velocity of the SAW and can be correlated to changes in the mass loading, viscosity and temperature of the substrate. The effect of the temperature on the substrate could be compensated by using a dual delay line configuration. In this case only one of the SAW devices will be functionalized with biological molecules. The reference SAW delay line is not functionalized with biological molecules and it is used only for temperature compensation. Both SAW delay lines will operate at the same temperature. Figure 11 illustrates a SAW biosensor in a dual delay line configuration (Arruda et al., 2009). Measurements comparing the experimental delay line with the reference delay line are used to compensate the effect of temperature on the biosensor.

A layer of bio-molecules consisting of protein cross-linkers and antibodies is coated on the sensitive surface of the sensing device in the path of the traveling waves, between the two sets of IDTs, as illustrated in Fig. 11. If specific target proteins (antigens) are present, they bind to the antibodies, creating the mass loading on the surface of the substrate. As a result, a time delays in the propagation of the SAWs will occur.

2.3 Guided Surface Acoustic Wave resonators

One subcategory of SAW, the Love propagation mode, is especially promising for chemical and biosensing applications due to its high sensitivity to mass loading and the ability to make measurements in liquid environments with minimal propagation losses. Shear-horizontal waves can be guided by placing a thin guiding layer on a SH-SAW sensor. The bare SH-SAW resonator has a lower sensitivity because the acoustic wave goes deeper into the substrate. The sensitivity to surface perturbations could be increased when a waveguiding layer fabricated on top of the IDTs is used. The waveguiding layer also provides protection from chemicals in the liquid. Waves that propagate through the guiding layer are known as Love waves. Dielectrics such as silicon dioxide, silicon nitride and most polymers can be used as waveguide materials. Polymers have a lower shear wave velocity and therefore they are recommended for Love mode SAW sensors. Acoustic efficiency is improved when the waveguide layer is thin and does not load or attenuate the traveling acoustic wave. A typical Love mode SAW sensor is illustrated in Fig. 12.

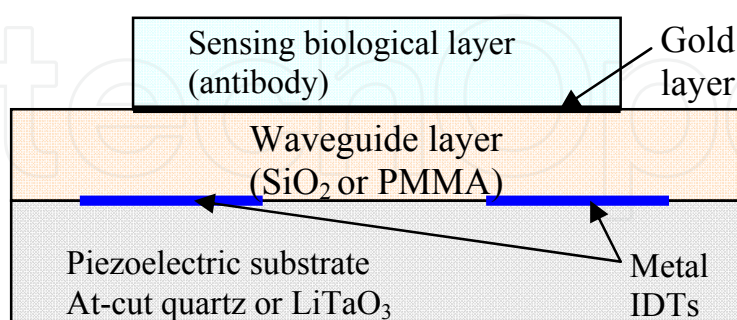


Fig. 12. Illustration of a Love mode surface acoustic wave sensor formed by a piezoelectric substrate, the waveguide layer, IDTs, and the sensing layer. The waveguide layer could be a SiO_2 film or a polymer layer and its function is to minimize propagation losses.

An example of a MEMS Love wave SAW device was implemented on LiTaO_3 substrates with Cr/Au IDTs and poly(methyl methacrylate) (PMMA) waveguide (Bender et al., 2000;

Branch et al., 2004). The waveguide is used to trap the SAW along the surface of the piezoelectric substrate to minimize energy losses and to protect the IDTs from corroding in a liquid-sensing environment. The surface of the waveguide is in contact with the sensing layer and it is often covered with a 50 nm thick gold layer, as illustrated in Fig.12. This gold layer provides better adherence of the sensing layer (antibodies) and prevents the nonspecific binding of proteins to the active area. Fabrication of the Cr/Au IDTs used standard lithographic techniques and the PMMA layer was spin-coated on the device. The thickness of the PMMA layer was optimized to avoid acoustic attenuation. These devices were tested with goat immunoglobulin G (IgG) and indicated minimum mass sensitivity of 17 pg/mm².

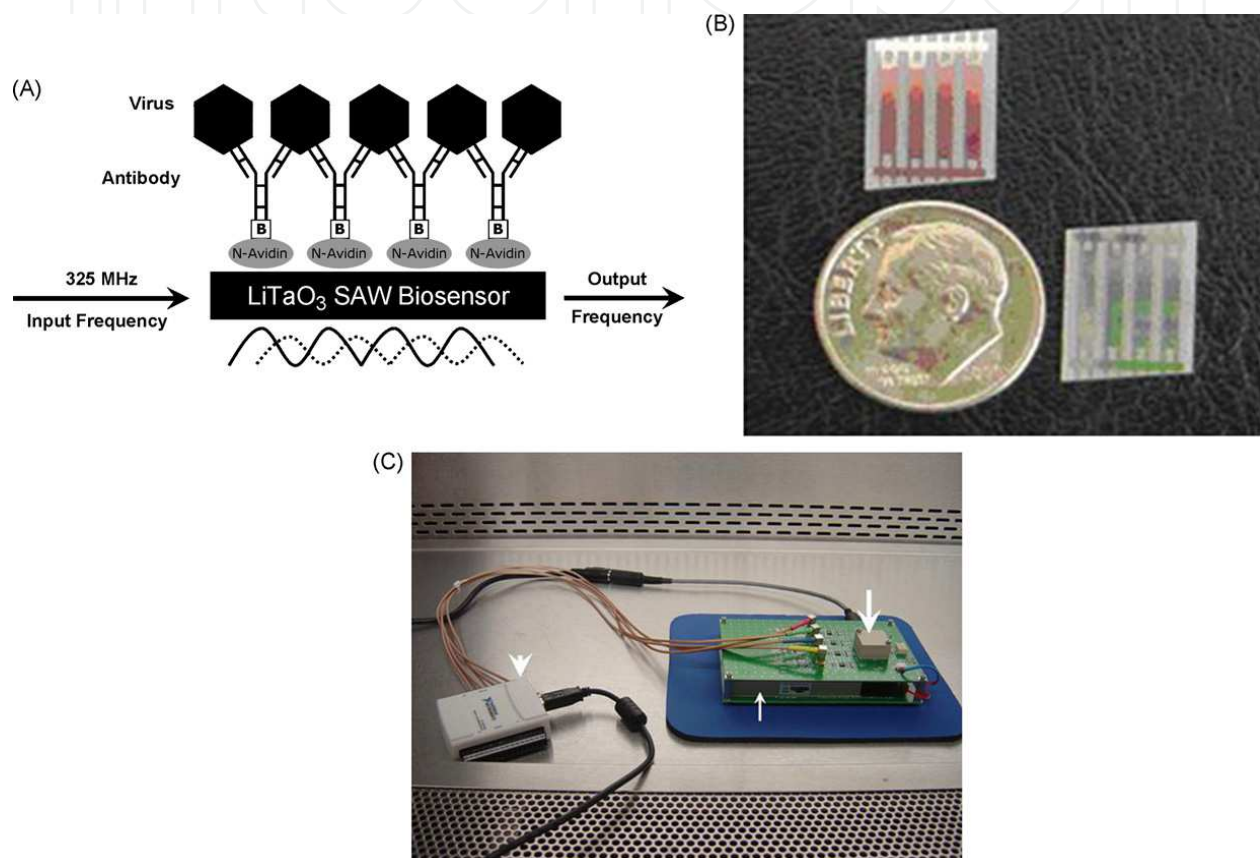


Fig. 13. (A) Concept of the antibody-based virus surface acoustic wave (SAW) biosensor. The lithium tantalate (LiTaO_3) sensor surface was coated with NeutrAvidin Biotin Binding Protein (N-Avidin) and coupled to biotinylated (B) monoclonal anti-JVB antibody for Coxsackie virus. Anti-SNV-G1 glycoprotein scFv antibody for SNV detection was coupled directly. Molecular interaction between virus and antibody elicits an acoustic wave leading to a change in the input frequency of 325 MHz. (B) The sensor wafer is shown in scale compared to a dime coin; four aluminum delay lines are visible; one serves as the reference and three as the test delay lines. (C) The SAW detection board (thin arrow) with the fluidic housing (thick arrow) and the output interface device (arrowhead) to a laptop computer is shown. Reprinted with permission from Ref. (Bisoffi et al., 2008). (© [2008] Biosensors and Bioelectronics). Used with permission.

Love wave SAW biosensors have also been used to test specific binding of different concentrations of Immunoglobulin G in the range of 0.7- 667 nM using a sensing surface

modified with protein A (Gizeli et al., 2003). In this case three different Love wave SAW resonators with different piezoelectric substrate were used: (1) a LiTaO_3 substrate operating at 104 MHz, (2) a quartz substrate operating at 108 MHz and (3) a quartz substrate operating at 155 MHz. This biosensor employs polymer waveguides fabricated using simple spin coating methods (Gizeli et al., 2003). Results indicate that the thickness of the polymer guiding layer is critical factor that determines the maximum sensitivity for a given geometry. It was also found that increasing the frequency of operation results in a further increase in the device sensitivity to protein detection.

A Love wave SAW sensor with a sensing layer of anti-bacillus antibodies was used to detect low levels of bacillus thuringiensis in aqueous conditions. Tests using bovine serum albumin (BSA) in place of *B. thuringiensis* spores indicated a detection limit of 0.187 ng BSA (Branch et al., 2004). When applied to the detection of bacteria in food and water, SAW biosensors allow rapid, real-time, and multiple analyses, with the additional advantages of their cost effectiveness and simplicity. The drawbacks associated with this kind of biosensors include relatively long incubation times of the bacterial sample on the biosensor surface, problems with crystal surface regeneration, high packaging cost, and difficulties to implement the related microfluidic system.

A lithium tantalate-based Love wave SAW transducer with silicon dioxide waveguide sensor platform featuring three test and one reference delay lines was used to adsorb antibodies directed against either Cocksackie virus B4 or the category A bioagent Sin Nombre virus (SNV), a member of the genus Hantavirus, family *Bunyaviridae*, negative-stranded RNA viruses (Bisoffi et al., 2008). The biosensor was fabricated using metal evaporation, plasma enhanced chemical vapor deposition (PECVD), and RIE techniques on a 36° y -cut, x -propagating lithium tantalate (LiTaO_3) wafer. This biosensor, illustrated in Fig. 13, was able to detect SNV at doses lower than the load of virus typically found in a human patient suffering from hantavirus cardiopulmonary syndrome (HCPS). Further, in a proof-of-principle real world application, this Love wave SAW biosensor was capable to selectively detect SNV agents in complex solutions, such as naturally occurring bodies of water (river, sewage) without analyte preprocessing.

3. Acoustic wave MEMS devices used for telecommunications

The explosive growth of the telecommunications industry in the recent decades has created a demand for high quality, compact and mobile radio frequency (RF) modules (Reindl et al., 1996). These mobile terminals typically consist of RF integrated circuits (RFICs) and a multitude of passive components. Driven by the success of the wireless technology business and marked progress in the submicron semiconductor fabrication techniques, acoustic wave technology has progressed to GHz range in recent years (Reindl et al., 1996). Acoustic wave devices in telecommunications are typically named according their acoustic wave propagation modes; bulk and surface. Bulk acoustic wave devices have the acoustic waves propagating through the substrate. Surface acoustic wave (SAW) resonators have the acoustic waves propagating along the top surface of the device. In this section we illustrate current activities in the acoustic wave resonators for both film and surface modes.

SAW devices are technologically mature compared to bulk wave devices. The invention of thin-film interdigital transducer (IDT) as a means of manipulating surface acoustic waves (SAW) in 1965 by White and Voltmer (Visser et al., 1989; White et al., 1965) has led to the

development of numerous complex electronic signal processing devices. This ingenious idea allowed a method of controlling electronic signals by transforming them into acoustic waves and manipulating the waves using patterns on piezoelectric substrates (Campbell, 1998; Morgan, 1985). Using the basic theoretical description of surface acoustic wave propagation presented by Lord Raleigh and the photolithographic techniques of microelectronics which are capable of fabricating small-size IDTs, a proliferation of radio-frequency (RF) SAW devices have been designed with innovative applications in wireless communications, radar and broadcasting systems (Hunter et al., 2002; Reindl et al., 1996; Weigel et al., 2002). Among examples of fabricated SAW devices are as components in satellite receivers, remote control units, keyless entry systems, television sets to identification tags (Campbell, 1998; Hikita et al., 2000; Springer et al., 1998). Other emerging applications of SAW resonators include gas sensors (Sadek et al., 2006), biosensors (Z. Xu, 2009), chemical (Nomura et al., 1998), temperature and pressure sensors (Buff et al., 1997). The significance of these devices in this industry can be measured in numbers by the worldwide production of these devices, where approximately 3 billion acoustic wave filters are used annually, primarily in mobile cell phones and stations (Reindl et al., 1996).

One of the most important applications of SAW resonators in telecommunication systems are as oscillators. In general, there are two major categories of MEMS resonators used in oscillator circuits namely 1) Purely mechanical resonator or flexural mode resonator can be formed using a beam or disk which will vibrate at a specific frequency and 2) Electro-mechanical resonators, usually formed using a piezoelectric membrane which is electrically excited to produce a traveling mechanical acoustic wave at a specific resonant frequency. The performance of the resonators is highly dependent both on the structure and design parameters of the resonator (Rebeiz, 2003). The essential requirements of a resonator with wireless applications include having precise resonant frequencies (f_r , f_p), low insertion losses, and high quality factors (Q) in the range of 10,000s (Zhou, 2009). Motivation of such devices is two-fold, they have powerful signal processing capabilities through control of surface waves and they are easily manufactured. Fabrication of SAW resonators typically requires a single deposition step, in comparison to their more fabrication-complex film bulk acoustic wave resonators counterparts (Ruby et al., 2001).

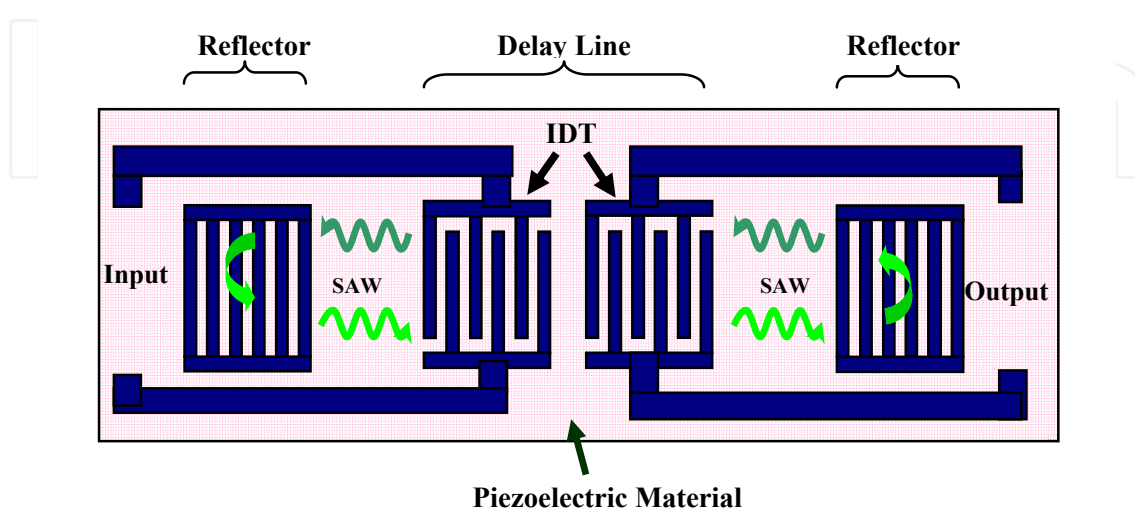


Fig. 14. Schematic of a Surface Acoustic Wave resonator

A typical SAW device (shown in Fig. 14) is composed of a piezoelectric substrate with thin-film metallic structures such as IDTs and reflectors deposited on top of the substrate's surface (Reindl et al., 1996). The operating principle of a SAW device is based on the piezoelectric effect where an applied microwave voltage input at the transmitting (input) IDT generates a propagating acoustic wave on the surface of the substrate (Campbell, 1998; Morgan, 1985; Reindl et al., 1996). This propagating acoustic wave in turn produces an electric field localized at the surface which can be detected and translated back into an electrical signal at the output IDT port (Morgan, 1985). Different from SAW delay-lines which operate based on traveling acoustic waves, SAW resonators operate using standing waves. Standing waves, or resonance, are created by the presence of reflectors, which contain the acoustic waves within the cavity. The array of metal strips or reflectors minimizes losses by reflecting and containing the acoustic waves within the cavity, thereby reducing the losses of the waves propagating outwards (Morgan, 1985).

SAW devices are typically fabricated on piezoelectric substrates and are packaged as discrete components. Very little silicon integration is achieved since these devices are connected to their CMOS counterparts on printed circuit boards, resulting in an overall large footprint of the system. To match the CMOS-based RF-circuitry and to realize a single-chip transceiver system, there have been efforts to integrate SAW devices on silicon. One such example is illustrated in (Nordin et al., 2007) where a CMOS SAW resonator was fabricated using 0.6 μm AMIs standard CMOS technology process with additional MEMS post-processing. The cross-section of the device is shown in Fig. 15. The SAW transducers are placed underneath the piezoelectric thin film to allow better CMOS compatibility. With this topology, the SAW transducers can be fabricated using the metal structures of the standard CMOS process. Thin metal wires of submicron width are common structures for fabrication of integrated circuits using the standard CMOS process. The MEMS SAW resonator can be placed beside the CMOS RF-circuits such as amplifiers to form an oscillator. Interconnections between the SAW resonator and the active circuits can be done using internal metal layers, reducing the parasitic effects of lossy, external bond wires and allows both the resonator and circuits to be placed on the same chip. This CMOS SAW resonator is still in the developmental stage and measured Q factors were less than 500.

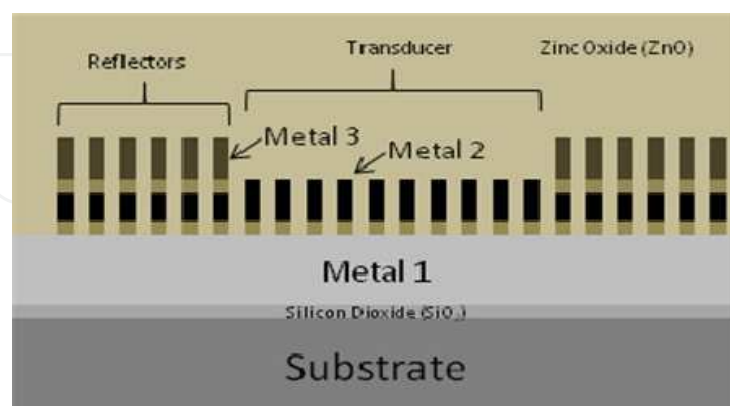


Fig. 15. Cross section of CMOS SAW resonator with increased height reflectors.

BAW resonators have acoustic waves propagating through the piezoelectric thin films. The metal transducers are placed on top and at the bottom of the piezoelectric material. Conductive metals such as Au, Mo or Pt can be used as electrodes. Acoustic waves can be

generated an electrical signal is placed at the top electrode and detected the bottom electrode. The thickness of the material determines the resonant frequency of the BAW device as shown in (1)

$$f_r = \frac{v}{2d} \quad (1)$$

where f_r is the resonant frequency, v is the acoustic wave velocity and d is the thickness of the piezoelectric material. Using (1), a thick piezoelectric layer will have low resonant frequency and vice-versa.

Integration with CMOS circuits for FBARs is more challenging compared to SAW resonators. The metal electrodes and piezoelectric thin films have to be deposited separately on silicon and cannot utilize existing CMOS processes. Interconnections between the FBAR and the CMOS circuits can be made using a separate metal deposition or using bond wires. However, FBARs are more attractive compared to SAWRs due to their high Q s (10,000) and low insertion losses. As such, a lot of work has been done to integrate silicon-based FBARs with CMOS circuits (Campanella et al., 2008; Hara et al., 2003; Otis et al., 2003; Zuo et al., 2010). Aluminum nitride is typically utilized as its piezoelectric layer due to its high coupling coefficient and silicon compatibility. An example of an AlN FBAR on Si is shown in Fig. 16 (Hara et al., 2003). Three different structures were fabricated in this work namely; the acoustic diaphragm type resonator, air gap type resonator (AGR) and solidity mounted type resonator (SMR). All three structures have acoustic isolation from the lossy silicon substrate to yield higher Q s. The AGR demonstrated superior experimental results of $Q = 780$ and an effective electromechanical coupling constant (k_{eff}) of 5.36 % compared to the other two structures.

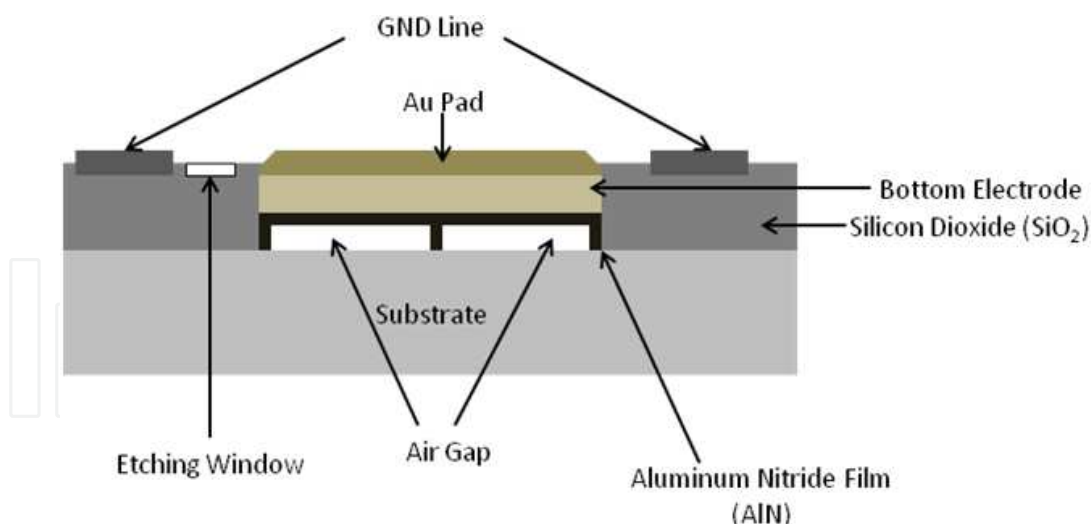


Fig. 16. Cross-section of a film bulk acoustic wave resonator

To improve the performance of the resonator without sacrificing silicon compatibility, hybrid SAW and BAW devices have been designed. An example of the hybrid device is the lateral-field excited (LFE) resonator, was successfully fabricated with Q factors in the order of 1000 (Zuo et al., 2010). This LFE resonator has Pt electrodes at the top and placed on Si substrates as shown in Fig. 17. The Si substrate is later back-etched to create the AlN LFE membrane. Elimination of the bottom FBAR electrode greatly relaxes the alignment

requirements. The resonant frequency of the LFE resonator is dependent on the periodic spacing of the IDT or λ similar to SAW devices. Measurements indicate high electromechanical coupling coefficient of 1.20%, due to the membrane structure and the highly conductive Pt electrodes.

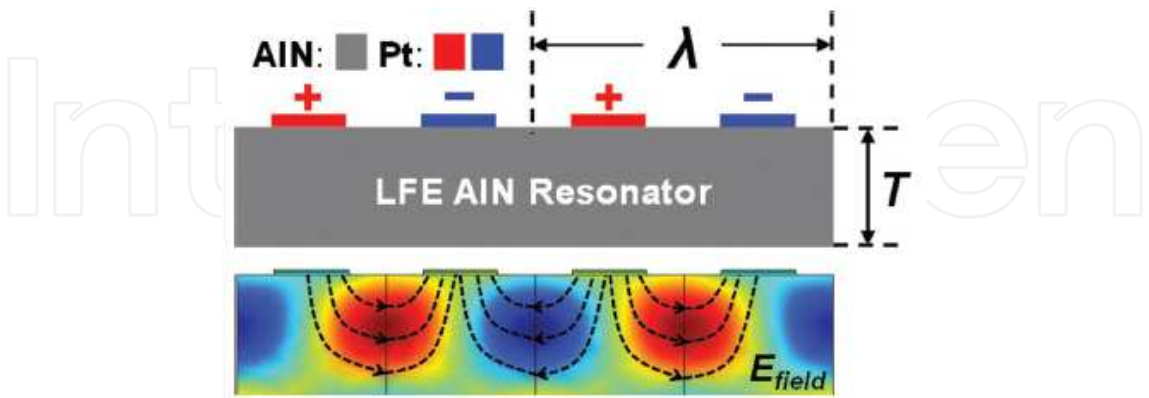


Fig. 17. Cross-section of a lateral-field-excited resonator (Zuo et al., 2010), (© [2010] IEEE). Used with permission.

Another hybrid SAW and FBAR device was reported in (Harrington et al.; Lavasani et al.) with even better Q of 6700 in air. The extensional thin-film piezoelectric-on-substrate (TPoS) shown in Fig. 18 uses molybdenum electrodes on top an AlN membrane, which is suspended above the Si substrate. The resonance frequency can be calculated using (2) as shown below:

$$2\pi f_s = (\pi/\lambda) \sqrt{(E_{eff} / \rho_{eff})} \tag{2}$$

Where E_{eff} and ρ_{eff} are the effectiveYoung’s Modulus and density of the composite structure, respectively. The AlN membrane is isolated from the lossy substrate, allowing the device to have very low motional resistance of 160 Ohms. The AlN membrane and Mo electrode is anchored to the substrate on both sides. Usage of multiple anchors were also investigated and has proven to suppress the spurious modes (Harrington et al.).

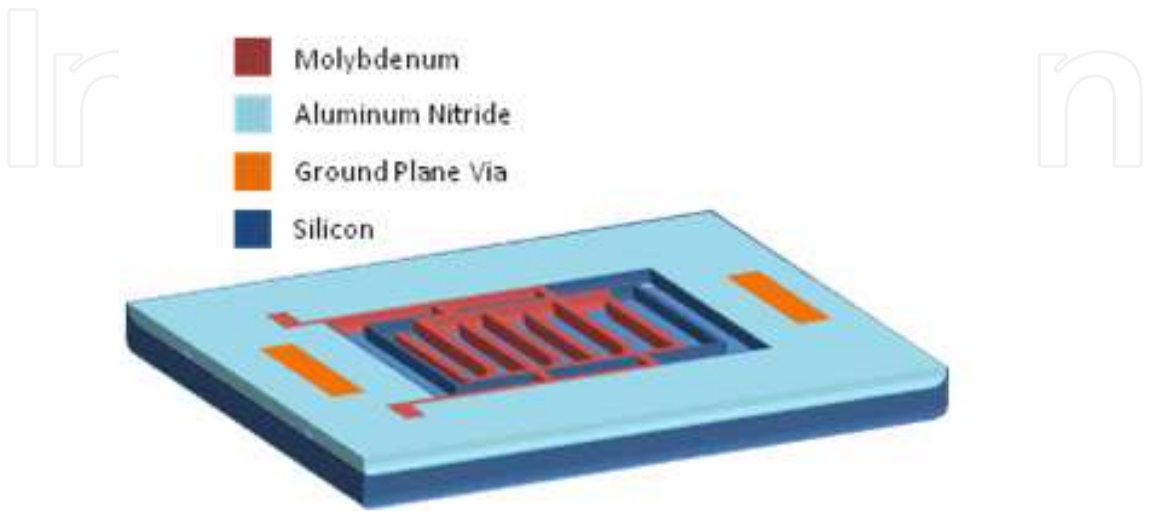


Fig. 18. Schematic for 7th order thin-film piezoelectric on substrate (TPoS) resonator.

4. Conclusion

This chapter is focused on two important applications of the acoustic-wave based MEMS devices; (1) biosensors and (2) telecommunications. Only few types of acoustic wave devices could be integrated in microfluidic systems without significant degradation of the quality factor. The acoustic wave based MEMS devices reported in the literature as biosensors are film bulk acoustic wave resonators (FBAR) and SAW resonators and delay lines. The Love mode SAW devices are often used as biosensor because the acoustic energy is confined to the sensing surface resulting in higher sensitivity to surface perturbations. The experimental results demonstrate that Love mode biosensors have high detection sensitivity.

Acoustic waves offer a promising technology platform for the development of biosensors and small-sized, low power RF-MEMS filters and resonators. MEMS acoustic wave biosensors are characterized by high sensitivity, small size and portability, fast responses, ruggedness and robustness, high accuracy, compatibility with integrated circuit (IC) technology, and excellent aging characteristics. Sensors based on this technology can be manufactured using standard photolithography and hence can be produced as relatively inexpensive devices. Integration of acoustic elements and electronic circuitry on a single silicon chip allows smart acoustic microsensors with advanced signal processing capabilities to be realized. Acoustic waves based biosensors offer the possibility of observing real-time binding events of proteins and other important biological molecules at relevant sensitivity levels and at low cost.

Acoustic wave MEMS devices used in telecommunications applications are also presented in this chapter. Telecommunication devices have different requirements compared to biosensors, where acoustic wave devices operating as a filter or resonator are expected to operate at high frequencies (GHz), have high quality factors and low insertion losses. Traditionally, SAW devices have been widely used in the telecommunications industry, however with advancement in lithographic techniques, FBARs are rapidly gaining popularity. FBARs have the advantage of meeting the stringent requirement of telecommunication industry of having Qs in the 10,000 range and silicon compatibility.

Currently, there is the concern that ZnO film, widely used as piezoelectric substrate for acoustic wave devices employed as biosensors and for telecommunications is very reactive, and unstable in liquid or air. Therefore, the stability and reliability of these devices become a problem. To solve this problem, the deposition of a thin protection layer such as Si₃N₄ on top of the ZnO film could be considered. Compared to ZnO, AlN shows a slightly lower piezoelectric coupling. However, AlN films have excellent piezoelectric properties. The Rayleigh wave phase velocity in (0 0 1) AlN is much higher than ZnO, which suggests that AlN is preferred for high frequency and high sensitivity applications (Gorla et al., 1999). AlN is a hard material with bulk hardness similar to quartz, and is chemically stable at temperatures less than about 700 °C. Therefore, using AlN could be an alternative and lead to the development of acoustic devices operating at higher frequencies, with improved sensitivity and performance in insertion loss and resistance in harsh environments (Mason et al., 1972)

The popularity of portable communication gadgets has increased the demand and necessity of small-sized, low power RF-MEMS filters and resonators. Passive acoustic wave resonators fulfill this market niche of low-power, radio frequency and silicon-compatible resonators and filters. Surface acoustic wave devices (filters and resonators) have long been

popular in the communications industry. To improve silicon compatibility, efforts have been made to implement the SAW resonator using standard CMOS process with minimal post-processing. Results indicate that while this device shows promise, significant improvement is required before the CMOS SAW resonator can meet the stringent communication requirements. In this aspect, FBARs have shown better performance in terms of quality factor (6000) and low insertion losses. However, complete CMOS-compatibility has not yet been achieved and the device still requires bond wires for connections to the circuitry.

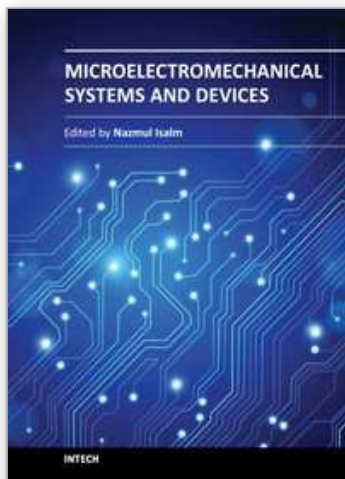
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The advances of microelectromechanical systems (MEMS) and devices have been instrumental in the demonstration of new devices and applications, and even in the creation of new fields of research and development: bioMEMS, actuators, microfluidic devices, RF and optical MEMS. Experience indicates a need for MEMS book covering these materials as well as the most important process steps in bulk micro-machining and modeling. We are very pleased to present this book that contains 18 chapters, written by the experts in the field of MEMS. These chapters are grouped into four broad sections of BioMEMS Devices, MEMS characterization and micromachining, RF and Optical MEMS, and MEMS based Actuators. The book starts with the emerging field of bioMEMS, including MEMS coil for retinal prostheses, DNA extraction by micro/bio-fluidics devices and acoustic biosensors. MEMS characterization, micromachining, macromodels, RF and Optical MEMS switches are discussed in next sections. The book concludes with the emphasis on MEMS based actuators.

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