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Design of MEMS Based Ultrasonic Transducer for Medical Imaging

[1]Chaitra M S, [2]Chethan M S
[1]Lecturer, [2]Assistant Professor
[1]Department of Electronics &Communication Engineering, [2]Department of Computer Science and Engineering
[1]Government Polytechnic Kartagi, Karnataka, India
[2]Shridevi Institute of Engineering and Technology Tumkur, Karnataka, India

Abstract: Ultrasonic medical imaging is a non-invasive diagnostic tool that uses high-frequency sound waves to visualize internal structures of the body. The design of a MEMS (Micro-Electro-Mechanical Systems) based ultrasonic transducer for medical imaging represents a significant advancement in this field, offering improved resolution, miniaturization, and compatibility with integrated circuits. This work presents the design and development of a MEMS-based ultrasonic transducer optimized for medical imaging applications. The transducer leverages piezoelectric materials and advanced fabrication techniques to achieve high sensitivity and wide bandwidth, enabling clear imaging with reduced noise and distortion. The compact and lightweight nature of the MEMS transducer facilitates ease of integration into portable and handheld imaging devices, making it ideal for point-of-care diagnostics. Through simulation and testing, the performance of the MEMS transducer is evaluated in terms of imaging quality, power consumption, and reliability. This work opens new possibilities for enhancing the accuracy and accessibility of ultrasonic medical imaging, ultimately improving patient care and outcomes.

***Index Terms* - Transducer, MEMS, Medical Imaging, GUI**

I. INTRODUCTION

Medical imaging is a field of study for scanning and visualizing human body for diagnosis. Imaging technology was used mainly for the defense and the space science communities in the past but its application has been expanded to medical field by the advent of powerful and less-expensive computers. The examples of medical imaging systems are x-ray, computed tomography (CT), magnetic resonance imaging (MRI), ultrasonic imaging, positron emission tomography (PET), etc. [1]

Each imaging system has advantages and disadvantages. For example, x-ray system is the most famous medical imaging equipment. The equipment is less expensive compared to other medical imaging systems such as CT, MRI, and PET. The system is also fast to get the result and the diagnostic procedures are simple. However, x-ray systems cannot be used for a pregnant woman and the image from x-ray system is only black and white. CT and MRI became available in the 1970s and 1980s, respectively, for advanced medical imaging. Physicians can obtain high-quality tomographic images of internal structure of the body and images with exceptional contrast for soft tissues from CT and MRI, respectively [2]. However, they both have some limitations. For example, CT cannot be used for a pregnant woman and MRI is not acceptable to a patient with metallic implant.

Ultrasonic imaging is the safest medical imaging system among the mentioned medical imaging systems. It is also very cost effective compared to CT, MRI and PET [3]. With reducing health care costs being a national agenda, this offers a unique advantage. It also has some other advantages such as real time imaging, higher resolution which may achieved by higher operating frequency, and the portability of the equipment. One major disadvantage is that the areas that can be scanned by ultrasound are still quite limited at this time. For example, it cannot be used to scan organs which contain gases. The quality of ultrasonic imaging also depends on the operator's skill [2].

This work is focused on ultrasonic medical imaging. The history of ultrasound as an imaging method dated back to late 1940s as part of the sonar and radar technology developed during World War II. It has evolved into a major diagnostic tool in medicine since early 1970s. The primary form of ultrasonic imaging is pulse-echo mode, and pulsed Doppler ultrasound devices also became available for measuring blood flow. Currently, ultrasonic imaging is the second most utilized diagnostic imaging system after x-ray. Even though ultrasonic imaging is a fairly mature tool in the areas of obstetrics, cardiology, and gynecology, its applications are still rapidly expanding with the newly developed technologies. Harmonic imaging, flow and tissue displacement imaging and multidimensional imaging are the results of advanced ultrasonic transducers [2]. Intravascular imaging with probes mounted on catheter tips at frequencies higher than 20MHz, endoscopic imaging with tip-mounted probes at frequencies from 5 to 20MHZ, and ophthalmological and dermatological imaging at frequencies higher than 50MHz are few examples of the results of the technologies for operating ultrasound at higher frequencies [3].The images generated by the current ultrasound technology are mostly two dimensional (2D) obtained by 1D array transducers combined with computed tomography [4-8]. However, any 3D reconstruction based on 2D images inherently reduces the available information because only the surface is shown or some depth related integral is performed [4].

A summary of the limitations of 2D images follows.

Volumetric information cannot be accurately determined. In 2D ultrasonic images, the volume is assumed to be approximately axisymmetric and 2D view is used as the basis for obtaining 3D images through extrapolation [4]. Thus, the calculated volumetric information based on 2D images is not as accurate as the information based on directly measured 3D ultrasonic images. The analysis of 2D ultrasound images is subjective because it relies on the experience and knowledge of the diagnostician for manipulating the ultrasound transducer, mentally transforming the 2D images into a 3D structure and making the diagnosis [5]. In quantitative perspective, 2D ultrasonic images are poor imaging method because it is difficult to localize the thin 2D ultrasound image plane in the organ and hard to reproduce a particular image location at a later time [5]. Since the location and orientation of 2D images are controlled by ultrasound transducer, sometimes particular views cannot be obtained because of the restrictions imposed by the patient's anatomy or position [6]. The limitations summarized above are expected to be overcome by 3D volumetric scan. Specifically, a real-time 3D ultrasound imaging offers the following unique advantages. 3D ultrasound images give more clear information about the scanned organ or tissue structure. Quantitative evaluation of 3D ultrasound images also reduces the subjectivity in the analysis of images and the results are less relevant to the diagnostician's experience [7]. Two ultrasound images are provided in Figure 1-1 to show the advantages of 3D ultrasound image compared to 2D image [2]. It is obvious that much more information can be retrieved from the 3D ultrasound image.

Real-time 3D ultrasound imaging does not require any form of post image manipulation on a computer to reconstruct 3D images from 2D images and it provides real volumetric images to allow 3D visualization of anatomy, the assessment of cardiac anatomy, and function during a single cardiac cycle [5-7].

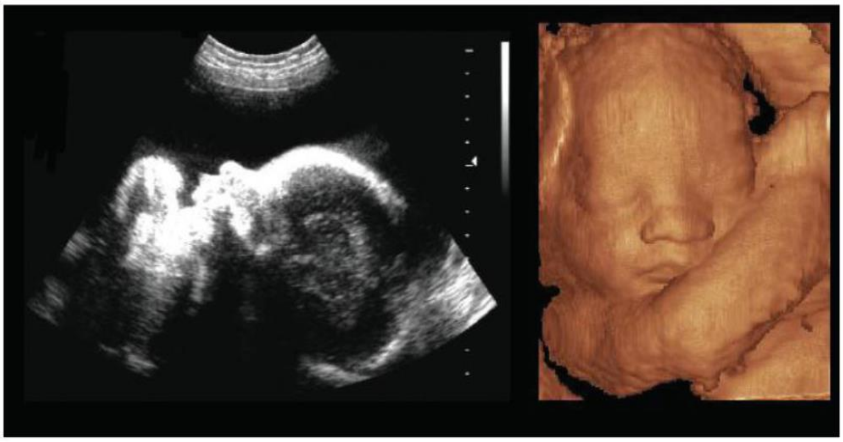

Fig1.1: 2D (left) and 3D (right) ultrasound images of a third-trimester fetus [2].

Ultrasound provides high frame rate (10 to 60 images per second) topographic images and the orientation of the images is flexible because they are not necessarily acquired as a stack of planes [5, 6]. The frame rate for CT and MRI is usually much slower compared to the frame rate of ultrasound and the orientation is fixed for CT and MRI [5, 6]. The high frame rate of image acquisition and the flexibility of the ultrasound provide the potential possibility in extending ultrasound imaging from its 2D to 3D and dynamic 3D (4D, real-time 3D) visualization. Figure 1-2 shows the comparison between an axial slice of 3D ultrasound and MRI images of infant heart. As can be seen in Figure 1-2, 3D ultrasound and MRI provide similar information about ventricle size and structure at this level [9].

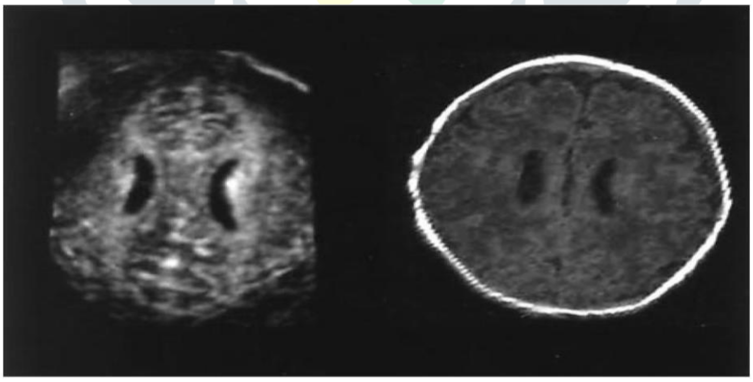

Fig 1.2: Axial slice of 3D ultrasound (left) and corresponding MRI (right) images of infant heart [9].

3D imaging also minimizes the probability to rescan the patient. Obtaining a single 3D scan of the entire region of interest will take less time than scanning the patient in real-time to search, record, and find the optimal 2D views [4]. Once 3D image is generated and saved, 2D cross-sectional image can be obtained in any orientation without restriction [6]. Examples of cross-sectional images from a volumetric 3D image are given in Figure 1.3 [2].

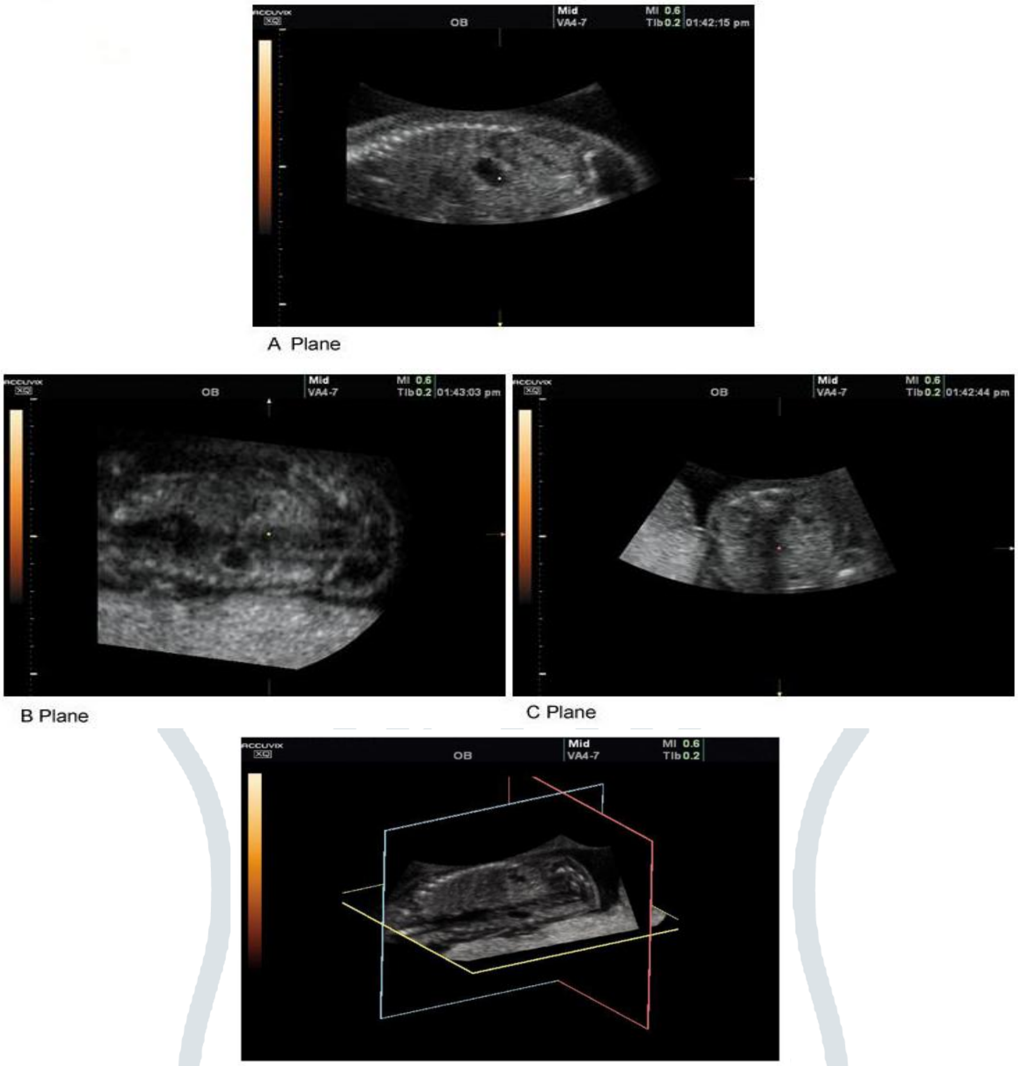

Fig1.3: Three cross sectional images (a) from a volumetric 3D image of abdomen of a seven-month fetus [2].

The real-time 3D images can be obtained by 2D arrays ultrasonic transducers [10, 11]. The main requirement of 2D array ultrasonic transducers for 3D imaging is the availability of transducers/arrays offering better resolution, sensitivity and its fabrication technique [3]. This project work is mainly focused on the design of micron-size ultrasound transducers using MEMS technology.

II. Related Work

Use of ultrasound in medicine started in the 1930s. Piezoelectric crystals (e.g., Rochelle salt and quartz) and magneto strictive materials (e.g., nickel) were the transduction material of choice until the 1940s. The intense materials research during World War II gave birth to the second generation of transduction materials, the piezoelectric ceramics (e.g., barium titanate and lead zirconate titanate). Electronic sector scanning for ultrasonic diagnosis was introduced in the late 1960s. The tensile piezoelectricity in stretched and poled films of polyvinylidene fluoride (PVDF), a polymer, was demonstrated in 1969. Linear arrays with electronic scanning started replacing fixed-focus mechanical sector scanners in the 1970s, providing greatly improved resolution and faster image formation. The details of the history of ultrasound imaging and transducer technologies outlined can be found in several papers [19]–[21].

Throughout the history of ultrasound imaging, piezoelectric crystals, ceramics, polymers, and recently piezocomposite materials [22] have been used to generate and detect ultrasound. Although the idea of capacitive ultrasound transducers is as old as the early piezoelectric transducers, piezoelectric materials have dominated ultrasonic transducer technology. The reason why capacitive transducers have not been popular is that electric field strengths on the order of a million volts per centimeter (106 V/cm) are required, so that electrostatic forces as large as a kilo- gram per square centimeter (kg/cm2) would be achieved, as the eminent French physicist Paul Langevin stated in 1915. However, recent advances in microfabrication technology have made it possible to build capacitive ultrasound transducers competing with piezoelectric transducers. Moreover, CMUTs offer advantages of improved bandwidth, ease of fabrication of large arrays with individual electrical connections, and integration with electronics [23]

Capacitive micromachined ultrasonic transducer (CMUTs) invented in the mid-1990s [24] have come a long way in the last two decades and recently reached the market for medical ultrasound imaging [25]. Considering the production of conventional ultrasonic transducer probes alone, which amounts to a global market of about $1billion annually, one can say that CMUTs can be the next big MEMS product in the medical field [26].

The concept of the ultrasonic transducer is similar to the condenser microphone (an electrostatic transducer). The MEMS based ultrasonic transducer consists of a suspended membrane, which is used to generate and detect the ultrasonic wave. The advantages of making the ultrasonic transducer in MEMS scale are the ability to generate and detect acoustic wave using only a single membrane structure, improvements in cost, and the high degree of reliability and performance [27]

In the early years of research in CMUT field, the main focus was on basic device fabrication and understanding device operation. Several fabrication processes based on standard surface micromachining techniques have been developed [28]-[32]. An alternative CMUT fabrication method based on wafer bonding was developed later [33]. Equivalent circuit models for CMUTs have been developed to help with the design of arrays for practical applications [14][34][35]. Finite element analysis (FEA) has been used to understand transducer characteristics, and to optimize transducer response [36]-[40]. 1D and 2D array elements have been fully characterized [41] [42]. Early imaging demonstrations were performed using systems built from discrete electronic components [43] [44].

III. BACKGROUND

The majority of conventional ultrasonic imaging systems are equipped with a variety of probes with conventional piezoelectric transducers that have a linear array of different frequencies. These arrays typically have been fabricated with the dice and fill technology, which is rather expensive and time-consuming [3]. Recently, due to the matured silicon micromachining technology, fabrication of micron size devices that are working at ultrasonic frequency range is possible [16]. Micromachined ultrasonic transducers (MUTs) are one application where MEMS miniaturization is expected to offer significant advantages over the current bulk piezoelectric transducers. Capacitive micromachined ultrasonic transducers (CMUTs) and piezoelectric micromachined ultrasonic transducers (PMUTs) that use MEMS technology have been introduced for array type transducer fabrication to overcome the limitations of conventional bulk piezoelectric transducers [17,18]. Since their advent, these two approaches have been the major working principles for developing novel transducers for ultrasound medical imaging applications. Compared to the traditional bulk PZT ceramic based ultrasonic transducers whose operating characteristics are controlled by the dimension and properties of the bulk PZT ceramic, the composite structures of CMUT and PMUT offer a much more flexible approach to the development of advanced micron size ultrasonic transducers. The performance of these miniaturized transducers has shown the strong possibility to replace the conventional ultrasonic transducers.

Real-time 3D ultrasound images promise many advantages compared to 2D ultrasound images as explained in Section 1.2. To get real-time 3D ultrasound images, 2D arrays of ultrasound transducers are necessary. Unfortunately, fabricating 2D array transducers by conventional dice and fill technology is almost impossible because of the enormous number of cables to be connected and the possible minimum transducer size. However, this restriction can be overcome by integrating MEMS technology into the array design and the fabrication of micron size ultrasound transducers. With the MEMS technology, those miniaturized transducers can be formed into a 2D array, which should be capable of producing real-time 3D images [3].Fabrication of thin film structures is both a time-consuming and costly process. Through the use of CAD for tools MEMS (Intellisuite, coventoware, comsol multiphysics), it is possible to simulate and virtually prototype devices before entering into expensive and time intensive fabrication. These simulation capabilities give designer the freedom to develop devices quickly within a digital environment.

IV. METHODOLOGY

In MEMS technology, CAD is defined as a tightly organized set of cooperating computer programs that enable the simulation of manufacturing processes, device operation, and packaged microsystem behavior in continuous sequence, by a microsystem engineer. CAD for Microsystems is still an emerging effort; a viable CAD package should include at least three major interactive databases: (1) electromechanical design database, (2) material database, and (3) fabrication database [55]. The content of a generic CAD package including the above database is schematically shown in figure 3.1.

As we can see from figure, the design database provides the necessary information and tools for design synthesis, codes for FEA (finite element analysis) and BEA (boundary element analysis), as well as tables and charts for other design considerations. The need for a material database in a CAD for Microsystems is obvious, as the properties of many materials used in Microsystems are not available from traditional material handbooks. This database should contain complete information on material properties. It should also include properties for transduction components such as piezoelectric and piezoresistive materials. The fabrication database, which is unique for microsystem design, involves all fabrication process simulations required for specifically selected fabrication and manufacturing processes as described in chapter 2. This database should also include wafer treatment such as the required cleaning processes for photolithography and thin film depositions. The results of these fabrication process simulations often include the inherent residual stresses and strains and other intrinsic stresses, which are used as input to the subsequent design analysis under normal operating and over-load conditions. Engineers can visualize the designed product in three dimensions by using the solid model option provided by CAD package. Most CAD packages have provisions for animations, which allow engineers to visualize the functions performed by the designed product [55].

The flowchart in the figure 3.1 is self-explanatory.Design engineers will first establish a ‘process table’ by selecting a substrate material once the product is configured from the design synthesis analysis. The CAD package will offer a possible PR photolithographic substrate treatment process from the fabrication process database. A mask is then either imported from external sources or created by the built-in design database for the subsequent photolithography on the substrate. The same database is then used to determine the appropriate fabrication process flow or steps that may include oxidation, diffusion, ion implantation, etching, deposition and other processes such as bonding, as selected by the designer. The CAD package offers detailed information on the selected processes, for instance, the etchants for the etching process with an estimated required time for each of such processes. The CAD package also provides automatic flow of information between the material database and the fabrication database. Once the fabrication processes have been established, electromechanical design begins. Here, the design engineer uses the solid model constructed by the CAD package for automatic mesh generation for the electromechanical analysis. Depending on the nature of the product, the CAD package can perform the finite element analysis for thermal conditions and mechanical strength of the structure, as well as electrostatic and electromagnetic analysis in the cases that involve actuation by the products. The later analysis requires the input of electrical potential and current to the finite element analysis. In addition to graphical displays for the analytical results, many CAD packages also offer animation of the designed product for kinematic and dynamic effects. Engineers may either terminate the design at this stage if the outcome of the design process is satisfactory, or make any necessary revisions to the configuration or loading or boundary constraints until all design objectives and criteria are met.

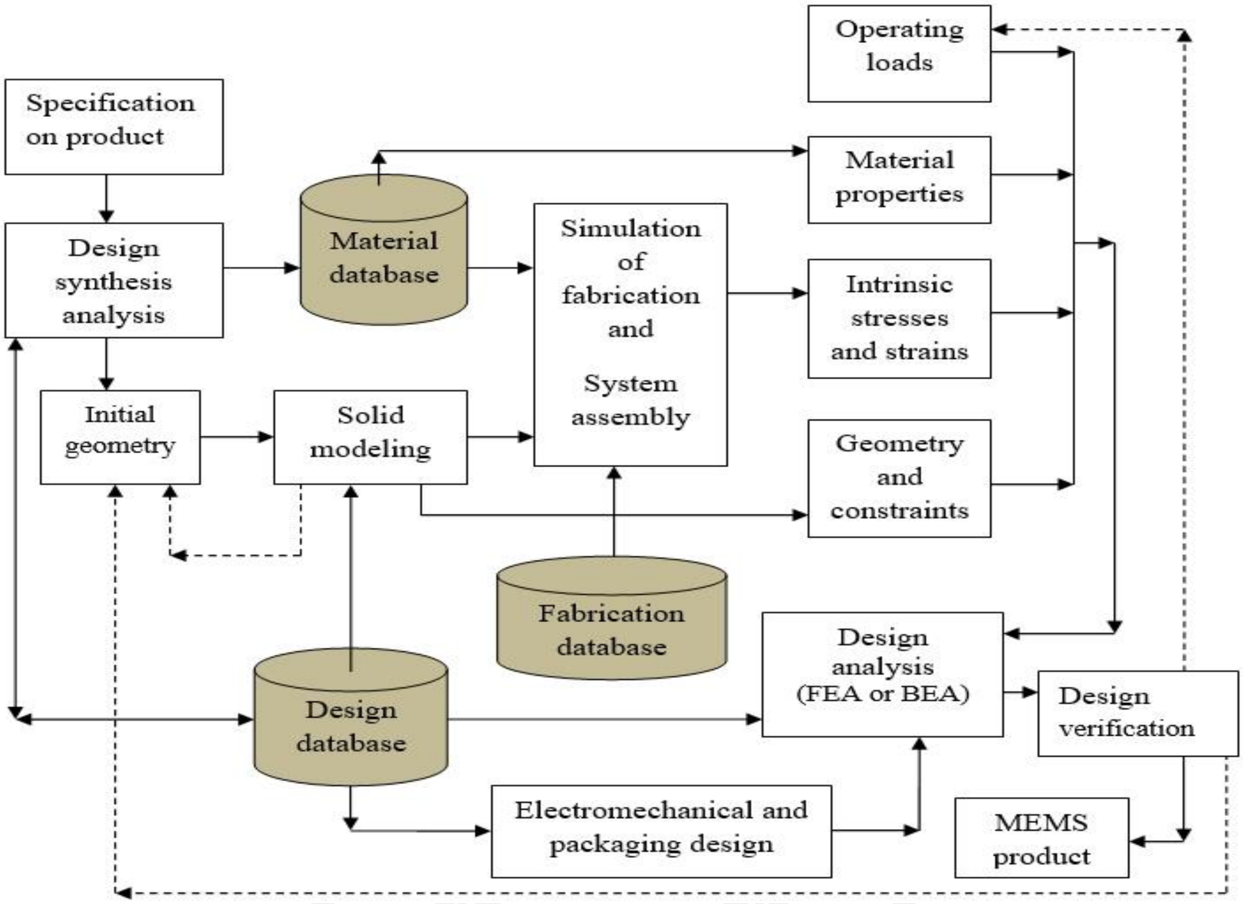

Fig 4.1 General structure of CAD for microsystem product design [55].

A. IntelliSuite

A commercial CAD tool with the trade name IntelliSuite is used in the current design of CMUT. It consists of three major databases similar to those shown in figure4.1

(1) Material database, (2) Electromechanical database and (3) Fabrication process database.

B. Process simulation

There are two methods to create the 3D models of MEMS devices in IntelliSuite; one is directly from the fabrication process, the other is through the 3D geometry interactive builder [56]. Using the fabrication process, the masks for the MEMS device were imported first, then a process table was generated which included all of the process steps necessary to create the device and from which the resulting material properties were determined. During process design, the imported mask set was linked to the process, which provided the definition of the x-y geometry of the structure. Then the 3D model of the device could be visualized in the 3D Viewer, and the model exported to an analysis module. Using the 3D geometry interactive builder to build a 3D model is like building with blocks. The layouts of the device should still be imported first. Each level, from bottom to top, was created by adding elements that described a final structure. The x-y geometry was obtained from the mask, and thickness (z) was designed by defining a level height. The different materials in device were defined by using multiple entities. Comparing these two methods of 3D model generation, the fabrication process is generally better if the designer knows the process. One advantage is that by using this approach the model will be easier to modify. The 3D interactive builder is useful when the process is not well understood or is difficult to define for simulation. It does allow for quick, simple model creation or the insertion of process steps, such as packaging using epoxy or solder [56].

C. Proposed Design

In the proposed design the CMUT of resonant frequency to be at least over 5MHz is considered, which can be used for diagnosing abdominal areas. Based on the outcome of the review, the proposed design presented in this work chooses the circular shape as the shape of a CMUT cell. The material and geometrical parameters are illustrated in table 4.1 and the cross section of a single CMUT cell is given in figure 4.2.

Layer	Material used	Thickness (µm)
Substrate	Silicon	10
Insulating layer	Silicon Nitride	0.1
Supporting wall (cavity gap)	Silicon Dioxide	1
Membrane	polysilicon	1

Table 4.1 Material and geometrical parameters

Fig 4.2 schematic of a single cell of CMUT

V. Results and Discussions

The Implementation is divided into two areas:

1) MATLAB Implementation
2) Development of Graphical User Interface

A. MATLAB Implementation

MATLAB was used to calculate various parameters like resonant frequency, collapse voltage, maximum displacement and comparative plots for different membrane materials were obtained. The influence of dimensional parameters on the collapse voltage and resonant frequency of the CMUT are analyzed. Three parameters of interest membrane thickness, membrane radius and cavity gap were taken into account in this study. These results can be used to design CMUT with specific collapse voltage and resonant frequency by modifying these parameters for different medical applications.

B. Effect of membrane Thickness on Resonant Frequency

The foremost parameter of ultrasound which decides its application in medical imaging is the resonant frequency. For example the table below shows resonant frequency of transducer for various medical imaging applications

Frequency (MHz)	Application
2.5	Deep abdomen, OB/Gyn
3.5	General abdomen, OB/Gyn
5.0	Vascular , Breast, Gyn
7.5	Brest, Thyroid
10.0	Breast , Thyroid , Superficial veins, superficial masses.

Table 5.1 Application of various ultrasound frequencies

The resonant frequencies are of particular interest in designing CMUT, because they indicate when the system will have its maximum response. It is given by the formula,

$$f = \frac{0.47 t_m}{a^2}\sqrt{\frac{y_0}{\rho(1-\sigma^2)}}$$

The critical parameters which influence the resonant frequency are the membrane thickness t_m , membrane radius a and the membrane densityρ. By varying the membrane thickness and membrane radius different resonant frequencies could be obtained for different membrane material like polysilicon, silicon and silicon nitride. The results obtained by implementing the above formula in MATLAB are presented below.

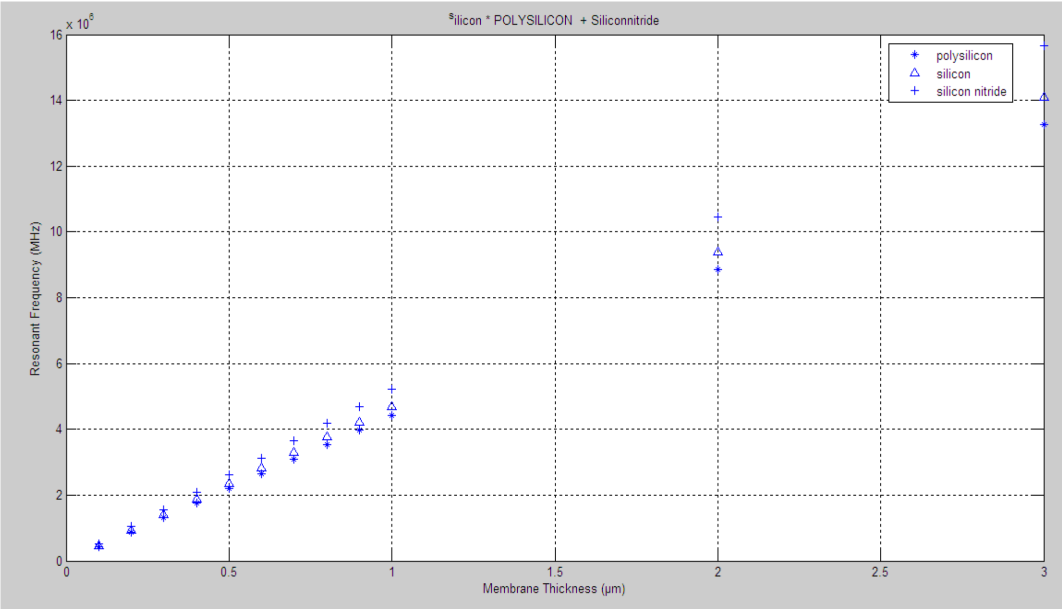

Fig 5.1 Membrane thickness Vs Resonant frequency

The resonant frequency were calculated for different combinations of membrane thickness (t_m= 0.1 to 3 µm) and membrane material (polysilicon, silicon, silicon nitride) is shown in figure 5.1 and table 5.2. The resonant frequency, corresponding to t_m=0.1 µm are 0.44MHz, 0.46 MHz, and 0.52 MHz for transducer with membrane material polysilicon, silicon and silicon nitride respectively. Similarly at t_m= 3 µm, the resonant frequency obtained for ploy silicon, silicon, silicon nitride are 13.2 MHz, 14MHz, 15.6MHz respectively. The result concludes that resonant frequency of CMUT is directly proportional to the thickness of the membrane; resonant frequency increases with membrane thickness. It is also observed that the optimal membrane thickness for medical imaging application is from 0.1 µm to 3 µm. The required ultrasound frequency range 1-15MHz could be obtained by varying the membrane thickness from 0.1 to 3 µm as shown in table 5.2.

Membrane Thickness (µm)	Resonant Frequency (Hz)		
	Polysilicon	Silicon	Silicon nitride
0.1	4.4223e+005	4.6974e+005	5.2219e+005
0.2	8.8446e+005	9.3947e+005	1.0444e+006
0.3	1.3267e+006	1.4092e+006	1.5666e+006
0.4	1.7689e+006	1.8789e+006	2.0888e+006
0.5	2.2111e+006	2.3487e+006	2.6110e+006
0.6	2.6534e+006	2.8184e+006	3.1332e+006
0.7	3.0956e+006	3.2881e+006	3.6554e+006
0.8	3.5378e+006	3.7579e+006	4.1776e+006

0.9	3.9801e+006	4.2276e+006	4.6997e+006
1	4.4223e+006	4.6974e+006	5.2219e+006
2	8.8446e+006	9.3947e+006	1.0444e+007
3	1.3267e+007	1.4092e+007	1.5666e+007

Table 5.2 Resonant Frequencies of CMUT with different membrane thickness

C. Effect of membrane radius on Resonant Frequency

The radius of the membrane ‘a’ has a wide impact in determining resonant frequency of CMUT. By varying the membrane radius different resonant frequencies could be obtained for different membrane material like ploysilicon, silicon and silicon nitride. It can be observed from the graph that the resonant frequency of CMUT is inversely proportional to membrane radius; resonant frequency decrease with increase in membrane radius.

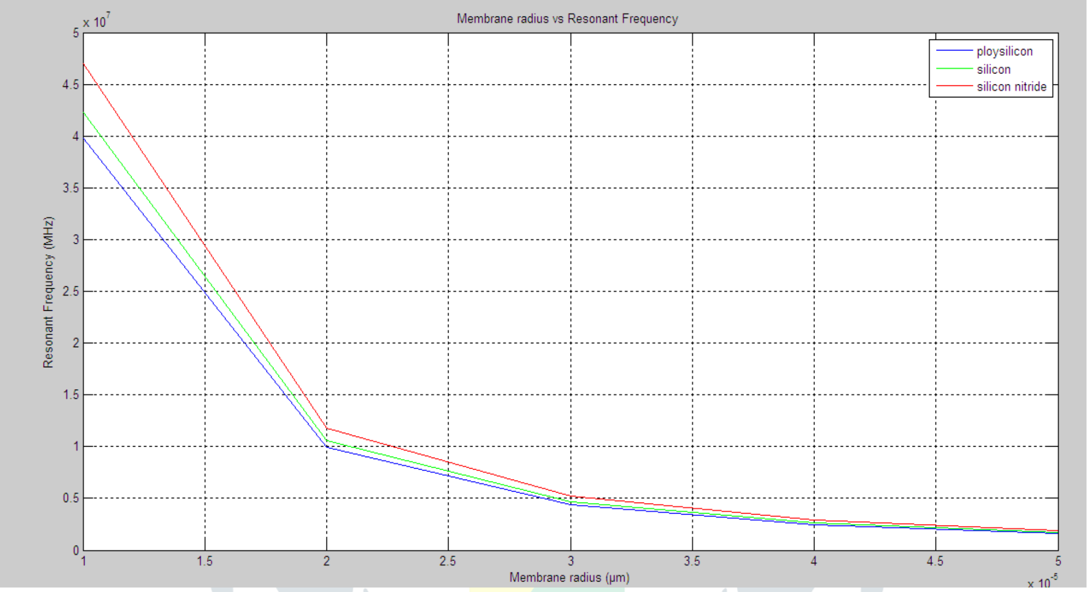

Fig 5.2 Membrane radius Vs Resonant frequency

Table 5.3 Resonant Frequencies of CMUT with different membrane radius

Membrane Radius (µm)	Resonant Frequency (Hz)		
	Polysilicon	Silicon	Silicon nitride
10	3.9801e+007	4.2276e+007	4.6997e+007
20	9.9501e+006	1.0569e+007	1.1749e+007
30	4.4223e+006	4.6974e+006	5.2219e+006
40	2.4875e+006	2.6423e+006	2.9373e+006
50	1.5920e+006	1.6910e+006	1.8799e+006

The resonant frequencies of transducer with three different membrane materials poysilicon, silicon, silicon nitride, corresponding to membrane radius a= 10 µm are 39.8MHz, 42.27MHz, 46.997MHz respectively. Similarly for a= 20 µm, 9.95MHZ, 10.5 MHZ, 11.7MHz. For a= 30 µm, 4.4 MHz, 4.6MHz, 5.2MHz. For a= 40 µm, 2.3MHz, 2.6MHz, 2.9MHZ. And for a=50MHz, 1.5MHz, 1.6MHz, 1.8MHz are obtained. This shows that the optimal membrane radius to design CMUT for medical imaging applications is between 10 to 50 µm.

D. Effect of Membrane Thickness on Maximum Displacement.

The maximum displacement of CMUT is given by,

$$x = \frac{1}{3}\left(t_a + \frac{\epsilon_0}{\epsilon}\, t_n\right)$$

Where, t_a= separation between the plates of CMUT
t_n= thickness of the insulator layer
ϵ_0= permittivity of free space
ϵ=$\epsilon_0\epsilon_n$, ϵ_n= relative permittivity of insulator layer.

The critical parameters which influence the displacement of membrane are cavity gap $t_a$and thickness of the insulator layer t_n. By varying the gap height different displacements could be obtained for different membrane material like ploysilicon, silicon and silicon nitride.

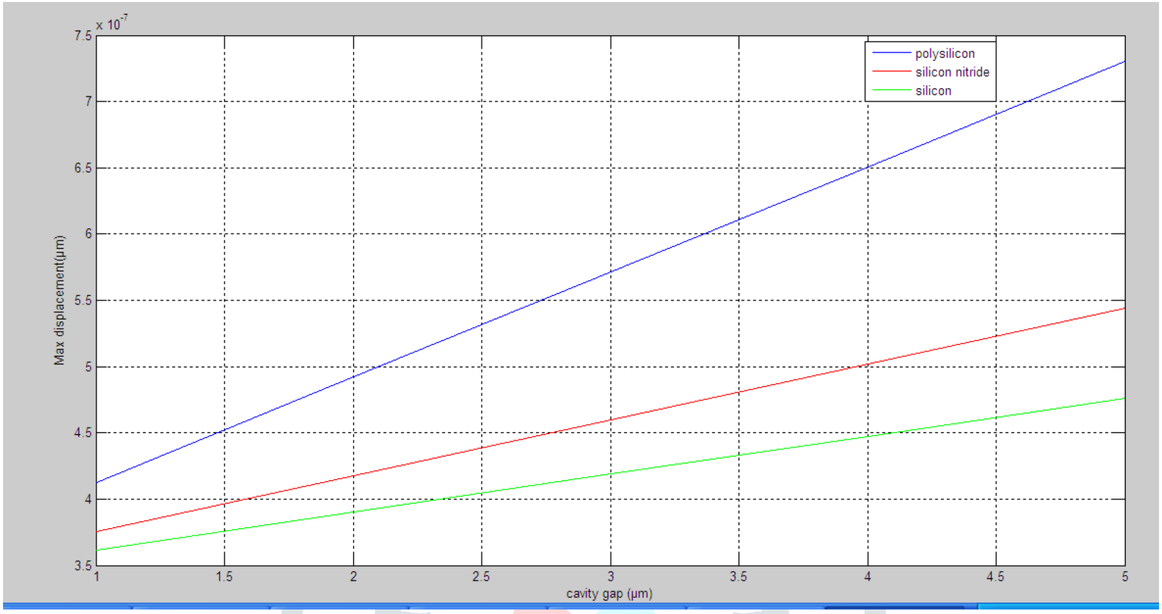

Fig5.3 Cavity gap height Vs maximum displacement

Cavity gap (µm)	Maximum Displacement (µm)		
	Polysilicon	Silicon	Silicon nitride
1	4.1270e-007	3.6182e-007	3.7553e-007
2	4.9206e-007	3.9031e-007	4.1772e-007
3	5.7143e-007	4.1880e-007	4.5992e-007
4	6.5079e-007	4.4729e-007	5.0211e-007
5	7.3016e-007	4.7578e-007	5.4430e-007

Table 5.4 Maximum displacement of CMUT with different cavity gap height.

E. Effect of membrane Thickness on Collapse Voltage.

As DC bias voltage is increased there is a point at which the electrostatic force overwhelms the restoring force of the membrane and membrane collapses. This particular voltage is known as collapse voltage. It is given by,

$$V_{collapse} = \sqrt{\frac{8k\left(t_a + \frac{\epsilon_0}{\epsilon} t_n\right)^3}{27A\,\epsilon_0}}$$

For a CMUT, collapse voltage is a critical parameter for employing the device at the optimum operating point. The operating DC bias voltage determines the performance of the transducer. It also determines the region at which the device is operating. The dimensional parameters which influence the collapse voltage are cavity gap height $t_a$and thickness of insultor layer t_n and area of the membrane A. by varying the different collapse voltage could be obtained for different membrane material like ploysilicon, silicon and silicon nitride.

F. GUI Database

This section discusses the various switches and displays in GUI that is created for interacting with the user. The GUI designed is as shown in the figure below. The GUI database includes the following. Membrane material panel: It contains various switches for different membrane materials on the GUI. Graphical plots panel: It contains the complete parametric analysis results and displays the comparative plots. Material property panel: It provides material properties to the user. User information panel: It provides the information regarding ultrasound frequency range used for different medical imaging applications to the user.

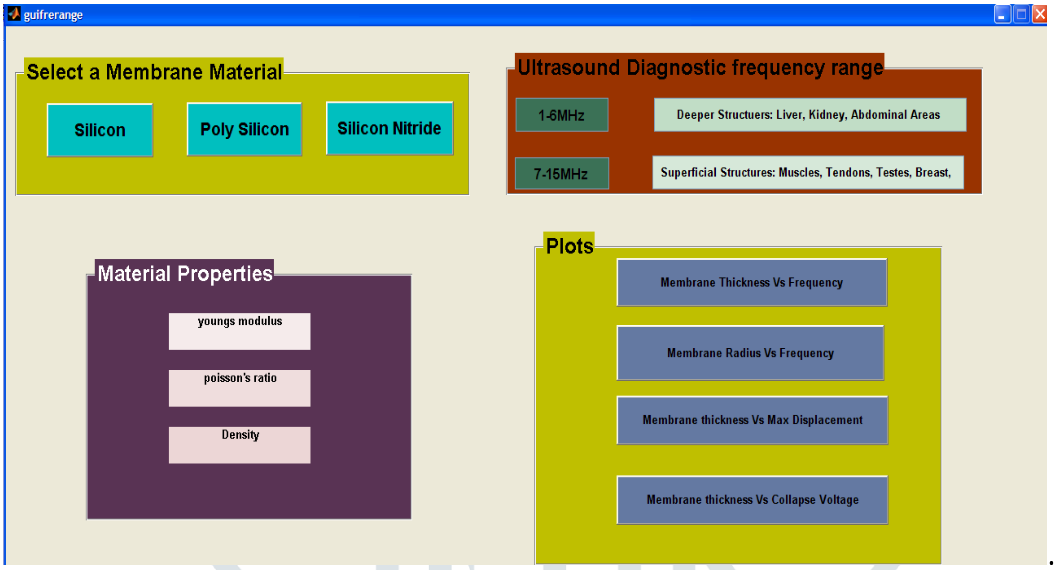

Fig 5.5 GUI Database

Often, a trial-and-error technique wastes both time and money when it is used to design devices. Therefore, developing a database which provides necessary information to design CMUT is desired. The designer can extract optimal parameters from this database before fabrication.

VI. CONCLUSIONS

Following conclusion are drawn from limited experimental investigation carried out in this project. Design and fabrication of single element CMUTs are the starting point for the fabrication of array CMUTs. Prior to optimizing array CMUTs, single element CMUTs must be characterized and optimized. The optimized design parameters of single element CMUT will be used for designing array type CMUTs.The designer can extract optimal parameters from this database before fabrication. A trial-and-error technique wastes both time and money when it is used to design devices.Improvements in the design of CMUT may be the focus of the future work

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