Capacitive Micro-machined Ultrasonic Transducer (CMUT) Based Volumetric Blood Flow-meter

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Abstract: In this paper a free membrane is used as a receiver to increase the capacitance and therefore the resolution of the flow meter. For the current application, from the wavelength of sound wave in soft tissue (c= 1540 m/s) the resolution was calculated to be 0.48 mm. This gives the first Eigen frequency of the capacitive structure according to which the poly silicon membrane was designed. After applying the bias, kinetic and dynamic study was carried out. Acoustic coupling gives the receiver mode. Changes were made to the dimensions to increase the resolution for the same fundamental frequency. The percentage increase in capacitance in the free membrane was calculated.

Keywords: CMUT, Ultra-sonic, MEMS, flow-meter

1. Introduction

According to the principle of imaging, we cannot resolve an object smaller than the wavelength of the measuring wave. Here, we have taken a resolution of 0.48 mm (for c= 1540 m/s in soft tissue). This implies that we cannot measure the volumetric flow-rate of arteries less than 0.48 mm from the surface of the probe which is well within our target area. This gives the frequency of acoustic wave as 3.32 MHz. The upper limit on velocity is limited by the Nyquist criteria. Taking this to be the first Eigen frequency of the membrane the capacitive structure was designed.
To improve the sensing of the vibration a free membrane was taken to increase (as opposed to bottom bound) the capacitance by a decrease in effective height.
The complete study is divided into the following parts:

1. Model design
2. Eigen frequency Analysis
3. Time dependent study
4. Stress/ Strain Analysis
5. Capacitance
6. Receiver mode

1.1 CMUT

CMUT consists of membranes with electrodes having A.C. plus D.C. bias. D.C. creates an electrostatic stress which is balanced by the spring force of the membrane. The alternating voltage varies the force generating the ultrasonic waves. These waves on striking the receiver generates an echo signal according to the capacitance change. The advantages of CMUT is improved bandwidth, ease of fabrication with silicon technology and integration with electronics[1]. Blood flow meter works on the Doppler shift in the echo signal to measure the velocity of blood stream. This device works in the time domain so that the ultrasound angle can be measured[2]. This also gives the diameter of flow and therefore the flow rate. It provides an edge over its laser counterpart since it can have a greater penetration depth owing to its low frequency when compared to laser making CMUT a minimally invasive device[1].

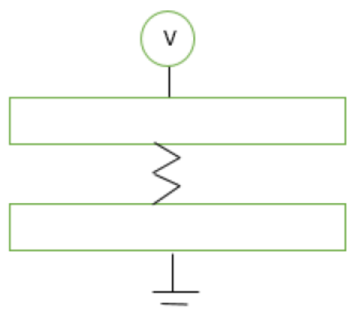

Figure 1: Lumped Model

1.2 Pulsed Wave Doppler measurement

Two pulses are transmitted with a recorded time delay. The distance moved is measured to find the velocity of the flow. In the time domain method maximum correlation method is employed to make sure that the two echoes are identical.[1]

The two pulses are separated by a time T with the respective echoes being received at T1 and T2 and the transducer axis makes an angle θ with the velocity flow. Therefore in time T, the distance moved by the fluid particle is cT1-cT2/2. From the fluid flow this distance is equal to VT cosθ. Therefore V= c (T2-T1)/2Tcosθ. If we use the sample with a time lapse of m periods the velocity V= c (T2-T1)/2mTcosθ.
The value T2-T1 is found at the position of maximum correlation. Since the maximum correlation will occur for θ=0, the correlation function will be a linear function of δT/m
The correlation is needed to find out the time delay as well the ultrasound angle. Theoretically it is the volume of overlap between the two beams. Analytically it is found out by integrating the kth pulse over the temporal offset.[1]

1.3 Resonant frequency calculation

The resonant frequency of a circular membrane can be determined from the elastic model of stiffness. Therefore,

$$fr = \frac{1}{2\pi}\sqrt{\frac{k}{m}} = \frac{tK}{2\pi a^2}\sqrt{\frac{E}{12\rho(1-\upsilon^2)}}$$

Where t is the membrane thickness, E is Young's modulus, ρ is the density, υ is the Poisson ratio and K depends on the anchoring.
As seen from the expression of resonant frequency above, the fundamental frequency depends on the t/a^2 ratio.
Also the center deformation is,

$$W_c = \frac{12pa^4(1-\upsilon^2)}{64Et^3} \propto (\frac{a^2}{t})^2 \times \frac{1}{t}$$

Keeping the same fundamental frequency for both transmitter and receiver, thicker a membrane is lesser center deformation it has. Therefore for better reception of ultrasound a thinner membrane with large change in capacitance is used. Decrease in *t* makes us reduce *a* as well for same fundamental frequency. For the transmitter a large acoustic pressure is needed which cannot be provided by a thin membrane therefore a thicker and larger membrane is used for transmitter.

3. Use of COMSOL Multiphysics

Structural shell mechanics module was implemented in COMSOL MULTIPHYSICS® on the model to find out the Eigen frequencies of the vibrating. A time dependent electro-mechanic study was set-up[3]. To implement a free membrane only the sides of the membrane was fixed. The poly silicon membranes were modeled as a linear elastic dielectric material and all three domains had an electrical material model. The air domain was free to deform. To simulate the receiver mode, the inward normal acceleration was coupled to the membrane acceleration and displacement amplitude was observed.

4. Computation

The A.C. plus D.C. biased membrane creates an electrostatic stress according to the Poisson's equation (1). The force density acting on the electrode comes from the Maxwell stress tensor (2)[3]. The varying force from the alternating voltage generates ultrasonic waves which waves on striking the receiver generates an echo signal according to the voltage change. Blood flow meter works on the principle of Doppler shift (3) in the echo signal to measure the velocity of blood stream. This device works in the time domain so that the ultrasound angle (4) (z is found experimentally) and the diameter of flow (5) can be measured which gives the flow rate.

$\nabla\cdot(\varepsilon\nabla V)=0$ .....(1)
Fes= -1/2[(E·D)n+(n·E)D] ....(2)
V= c (T2-T1)/2Tcosθ .….(3)
Tan θ=4zr /cδT …..(4)
D = cTsinθ/2 ..….(5)

4. Model Design

A circular polysilicon membrane with an air gap was modelled with the top and bottom boundaries as electrodes for voltage bias.
The figure below shows the rendering of the solid model.

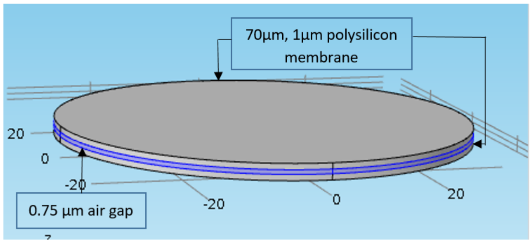

Figure 2: Solid model

Table 1: Model specifications

Parameter	Specification
Bottom membrane	Polysilicon
Radius	35µm
Thickness	1 µm
Air gap height	0.75 µm
Top membrane	Polysilicon
Radius	35 µm
Thickness	1 µm

Table 2: Polysilicon properties

Property	Value
Relative permittivity	4.5
Young's modulus	169GPa
Density	2320 kg/m^3
Poisson's ratio	0.22
Reference resistivity	$2*10^{-5}$Ω.m
Resistivity temperature co-efficient	$2*10^{-3}$ (1/K)
Reference temperature	298.15 K

Table 3: Air Properties

Property	Value
Relative permittivity	1

4. Eigen frequency Analysis

Structural shell mechanics was implemented to find out the Eigen frequencies of the vibrating membrane to account for the bending stiffness of the membrane.
The results are as follows:

Table 4: Eigen frequencies

Mode of vibration	Eigen frequency
1st mode	3.32 MHz
2nd mode	6.87 MHz
3rd mode	6.87 MHz
4th mode	11.18 MHz

Note that the 2nd and the 3rd Eigen frequencies are the same since the 3rd mode shape is equivalent to the 2nd mode shape rotated by 90◦.
Mode shapes

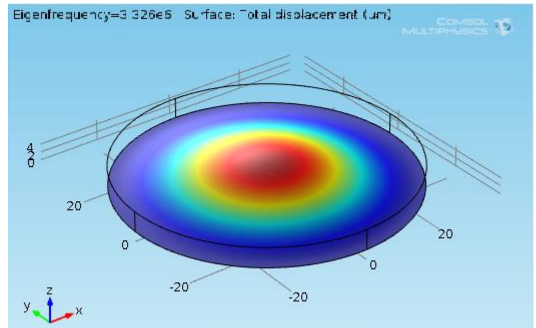

Figure 3: 1st resonant mode shape

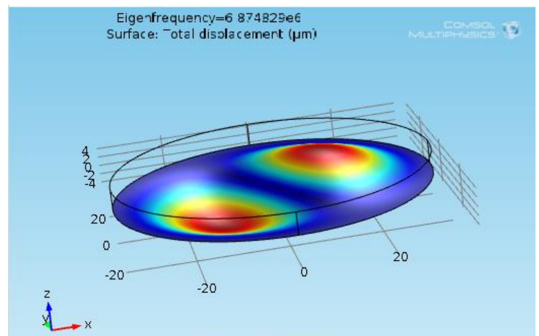

Figure 4: 2nd resonant mode shape

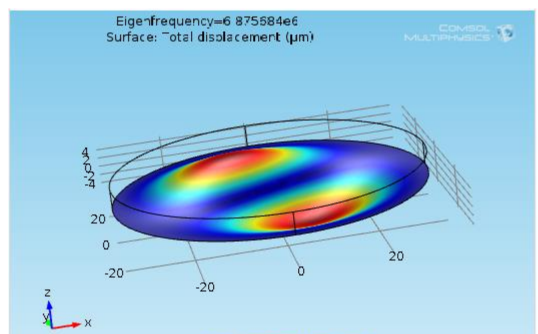

Figure 5: 3rd resonant mode shape

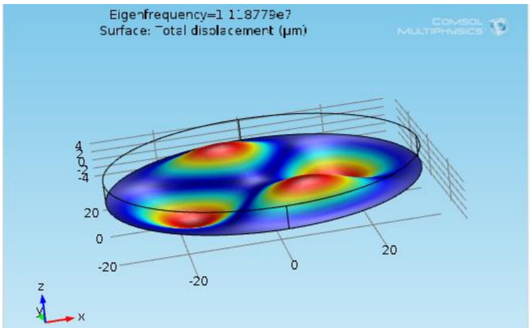

Figure 6: 4th resonant mode shape

5. Time dependent study

Taking the 1st Eigen frequency to be the principle mode of vibration a time dependent electro mechanic study was set-up.
The following shows the variable list for the study:

Table 5: List of variables

Name	Expression	Unit	Description
f	1.66e6[Hz]	Hz	
w	2*pi*f	Hz	
Vdc	135[V]	V	
Vac	15*sin(w*t)[V]	V	
Volt	Vdc+Vac	V	

Here the vibration frequency will be twice the bias frequency since the force is related to the square of voltage (shown in graph below). Therefore, f is taken to be 1.6 MHz.

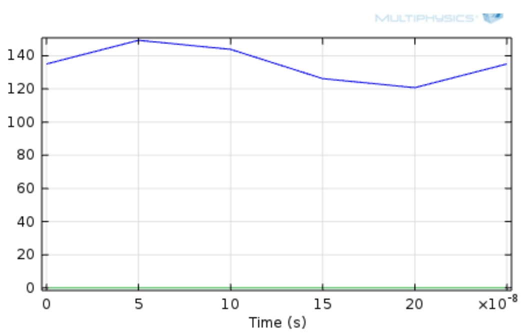

Figure 7: Electrostatic force completing a period in 1/f time.

Where the bias voltage 'volt' was applied to the upper membrane (shown below) and the lower membrane was grounded.

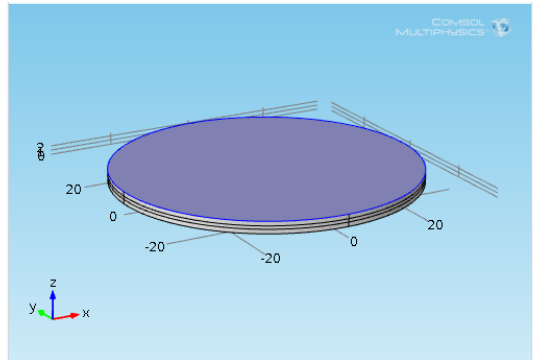

Figure 8: Biased top membrane

To implement a free membrane only the sides of the membrane was fixed while the bottom was free to move. This is shown in the following figure.

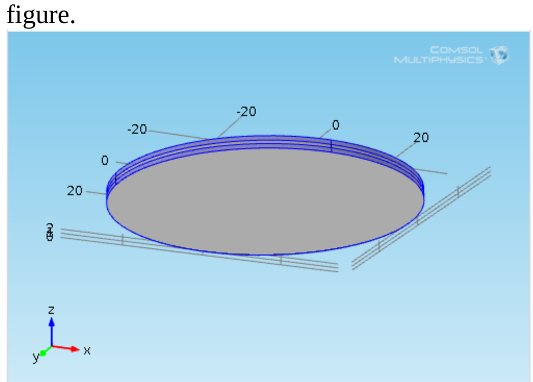

Figure 9: Fixed constraint of the membrane

The polysilicon membranes were modelled as a linear elastic dielectric material and all three domains had an electrical material model. The air domain was free to deform.
Displacement

The model showed a displacement amplitude of 0.29 microns. The figure below shows the displacement at different instants of time.

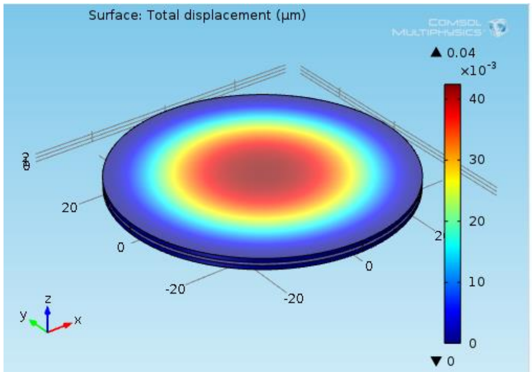

Figure 10: Displacement surface

The above result is scaled appropriately (scale factor 30) to help visualise the displacement shown in figure below.

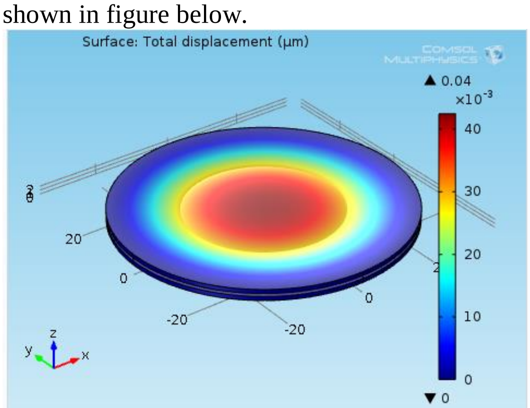

Figure 11: Scaled displacement surface

Notice here that since both the membranes vibrate inwards the scaled model shows them crossing each other. Actual displacement is less than the gap height.

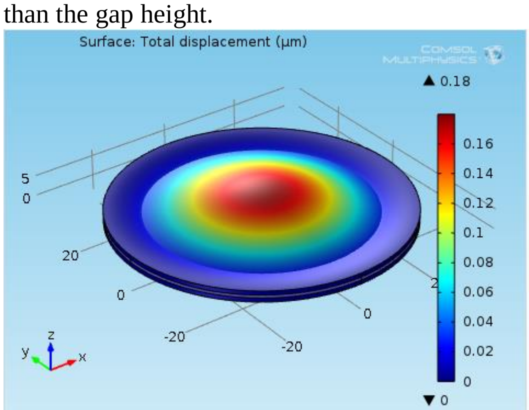

Figure 12: Displacement at an intermediate time

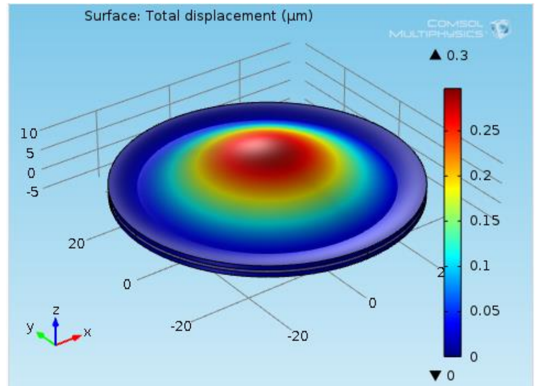

Figure 13: Displacement amplitude

6. Stress-strain analysis

The figure below shows the stresses developed in the polysilicon membrane. We notice a stress concentration at the edges raising the average stress value. This can be worked upon to reduce the stress.

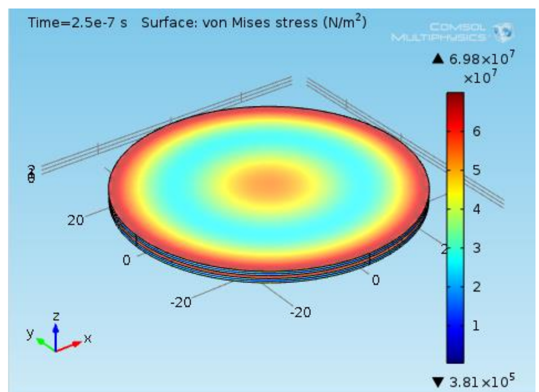

Figure 142: Von mises stress amplitude

To study the safe limit for the membrane, the first principle strain was measured against the maximum allowable tensile strain and the fracture strain.

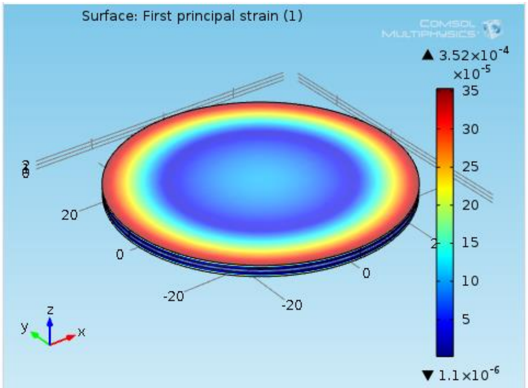

Figure 153: First principle strain

The strain developed was measured against the maximum allowable strain of LPCVD polysilicon equal to 0.018. This was found to be well within acceptable limits.

The z-component acceleration was computed to carry out an acoustic study of the device to check the acoustic intensity in the receiver mode.

Table 6: Acceleration at few instants of time

Time (s)	Acceleration, Z component (m/s^2)
0	-1.1972961431149285E7
5.0E-8	-1.1972016184040923E7
1.0E-7	-5153192.570307183
1.5E-7	1.5214310753995202E7
2.0E-7	1.4297464245830195E7
2.5E-7	4028475.9192867684

7. Capacitance

The capacitance values at DC bias was calculated when the membrane stresses inwards at its equilibrium position.
The figure below shows the increase in capacitance with a bound and free lower membrane. Here we see an increase of 6%.

Table 7: Capacitance values at DC bias

Capacitance (F)	Capacitance (F)
2.935599528928164E-14	3.060542933406574E-14

8. Receiver mode

To simulate the receiver mode, the inward normal acceleration was coupled to the membrane acceleration and displacement amplitude was observed.
The model showed a displacement amplitude of 0.23 μm which is close to that of the transmitter (0.29μm) at the same frequency of 3.2MHz

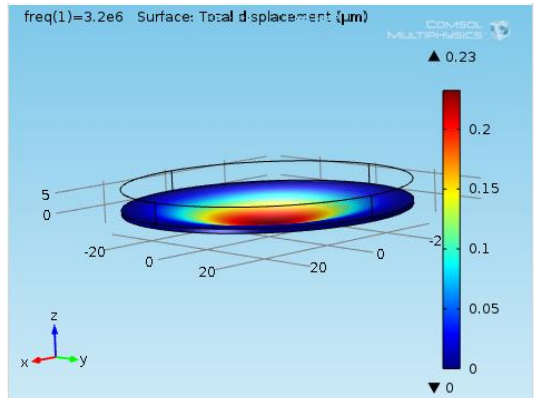

Figure 16: Displacement amplitude of the receiver mode

However, the receiving mode should have a much larger deformation so that distinct echo signals are obtained. Therefore, for the same fundamental frequency of 3.2MHz we can change the h/r^2 ratio to increase the center deformation by decreasing h.
Therefore, new dimensions are,
h=0.5μm
r= 24.75 μm
This gives us an amplitude of 0.46 μm.

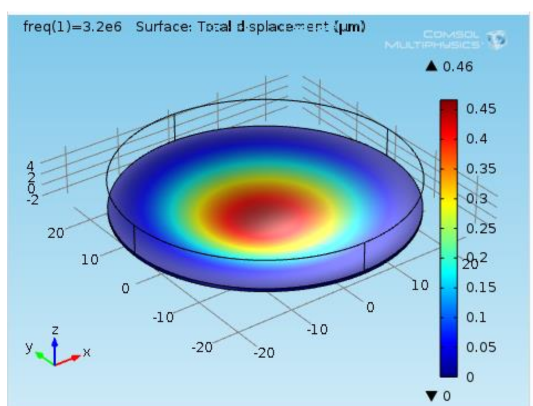

Figure 17: Modified displacement

9. Conclusions

The device is now proven to be more sensitive and is now ready for process modelling. Owing to the current silicon technology, this MEMS device can be easily mounted on a probe. Being minimally invasive it can be extremely useful for measuring the fetal cardiac blood flow. However, CMUT is a low frequency device and to further increase the resolution the device needs to be introduced inside the to-be measured body.

10. References

1) Wang, Mengli, CMUT arrays for blood flow ultrasound Doppler and photoacoustic imaging applications, Dissertation, University of New Mexico Albuquerque, New Mexico, December 2010

2) Wang, Mengli; Chen, Jingkuang, Volumetric Flow Measurement Using an Implantable CMUT Array, IEEE TRANSACTIONS ON BIOMEDICAL CIRCUITS AND SYSTEMS, VOL. 5, NO. 3, JUNE 2011

3) Electrostatically Actuated Cantilever, COMSOL Documentation

11. Acknowledgements

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