Microcantilever-based Biodetection

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Presented to:
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Abstract

This paper studies a microcantilever based detection method for application in biodetection, clinical diagnosis, and biochemical study. We aim to develop a cost-effective, portable platform for biodetection. For achieving this goal, a thorough literature review is presented. Based on this literature review and our final goal, a frequency sensing mode is chosen. Furthermore, the use of a piezoresistor as an output device is explored with the use of an external piezoelectric element as an actuation method. Following this, ANSYS software simulation is carried out to obtain an optimized geometry for the microcantilever, taking sensitivity and capture area into account. A trapezoid-like cantilever is found to be the best choice. Additionally, the possibility of sensing in higher frequency modes is investigated and 2nd order resonance frequency is chosen as the best. With the combined optimization of these parameters, a sensitivity of $1.05 \times 10^{18} \text{s}^{-1}\text{kg}^{-1}$ is achieved. Finally, a detailed microfabrication process together with mask design is given to push our device into reality. This process is fairly simple due to the utilization of a SOI wafer.
# TABLE OF CONTENTS

I. Introduction  ................................................................. 4

II. Literature Review .......................................................... 5

III. Design and Optimization ............................................... 8

IV. Fabrication ................................................................. 13

IV. Conclusion ................................................................. 17

Acknowledgements ............................................................ 18

References ................................................................. 18

Biography ................................................................. 20

Delegation of tasks .......................................................... 20

Appendix ................................................................. 21
I. Introduction

The fast development of micro-electro-mechanical system (MEMS) technology has brought many great ideas closer to reality, such as “lab on chip” technology [1]. The basic idea of “lab on chip” is to integrate all the functions of a bio-chemistry lab, such as sample preparation, mixing, detection, and data processing, into a single microchip. This promises to reduce workload, reaction time, and required sample volume due to scaling, enabling portable devices. One potential application of this technology would be to make HIV testing more readily available to developing nations, such as Africa, in which testing is currently limited to traditional, less effective methods. MEMS can provide microfluidic networks and pumps and mixers for handling sample fluids, as well as state-of-the-art detecting mechanisms utilizing piezoresistive and piezoelectric techniques. Among all these technologies, the microcantilever is the simplest MEMS based device having wide applications in biodetection. The theory that biomolecular interactions can cause microcantilever motion has been proposed by Oak Ridge National Laboratory [2], which speeds up the development of using microcantilever for biodetection. Several groups have demonstrated the ability of using microcantilevers to diagnose prostate cancer [3], myocardial infraction [4], and glucose monitoring [5].

The principle of microcantilever based sensing falls into two categories: static sensing and vibrational sensing. In static sensing, microcantilevers are usually deposited with a gold film which is further coated with a certain receptor. Upon binding the specific analyte (e.g. protein), stress is generated resulting in bending of the microcantilever. Piezoresistive or optical methods are used to read out the deflection and thus detect the targeting bioparticles.

In vibrational sensing, an external actuation method is required to vibrate the microcantilever through a certain range, and a sensing mechanism is also required to read out the amplitude of the microcantilever. From the frequency response spectrum, the resonant frequency is obtained. When certain bioparticles are adsorbed on a microcantilever, the resonant frequency will decrease due to the increased mass. This resultant decrease in frequency is measurable, providing a method by which the presence of specific bioparticles can be confirmed.

Our device is portable, as opposed to complicated and expensive optical equipment. It can be a cost-effective, portable platform for biodetection. In this paper, a thorough literature review on the vibrating sensing of microcantilever is presented. Optimization is done using ANSYS software to obtain an optimized geometry for the microcantilever taking both sensitivity and capture area into consideration. Furthermore, sensing in higher frequency modes is studied. Additionally, the microfabrication process and the mask design are presented with detailed process parameters.
II. Literature Review

The key elements to a microcantilever’s detection of biomolecules are the changes in the vibrational frequency or by measuring the micronatiliever’s deflection. In both cases biomolecules are detected when they are absorbed to the surface of a microcantilever. The surface of the microcantilever is coated with a special receptor that absorbs biomolecules.[6] After surface absorption, the microcantilever deflects several nanometers, because of its new mass. The deflection of the microcantilever is proportional to the concentration of the biomolecule [6].

We can determine the mass of the attached bio-molecule when we consider the vibrational frequency of a microcantilever. Again, the surface of the microcantilever is coated with a special receptor that will attract the biomolecule. The microcantilever has its own natural frequency \( \omega_R = (k/m)^{1/2} \), where \( k \) is the spring constant and \( m \) represents mass.[7] When a biomolecule attaches to the microcantilever, it changes the mass and the spring constant of the microcantilever. This in turn affects the resonant frequency of the microcantilever.[7] We can use that change in resonance frequency to determine the mass that has been attached to the microcantilever [7].

In order to measure the deflection of the microcantilever or the change in resonance frequency, we must choose a particular method. The methods that we shall discuss here are the optical method, the piezoresistive method, and finally the piezoelectric method.

The optical method requires the use of a low power laser beam that does not affect the biomolecules coated on the surface and the position sensitive detector (PSD).[8] If the microcantilever does not deflect then that means that no biomolecules have been absorbed by the microcantilever. In this case the laser beam would fall on a particular spot on the PSD. As the cantilever deflects, the position of the beam changes, which, in turn, is calculated using appropriate electronics. The strength of this method is that it can detect deflection to the sub-nanometer range. Major weaknesses include extraneous readings because of the heat from the focused laser beam, and high costs [8].

Piezoelectric microcantilever-based sensors detect the change in the resonance frequency of a microcantilever in order to determine the mass of the absorbed biomolecule [9]. In general this sensing is obtained through a thin-film plate which has a microactuator to drive the plate into resonance and a microsensor to determine the resonant frequency of the plate [13].

Piezoresistive microcantilever-based sensors measure the strain induced resistance change produced in the cantilever upon absorption of a particular biomolecule. In the embedded piezoresistive microcantilever (EPM) design, the cantilever is fully or partially embedded into this “sensing material” [6]. When the biomolecules are absorbed, the sensing material selectively adsorbs the foreign material, resulting in a tiny volumetric change in the sensing material. This volumetric change is measured as a simple resistance change in the piezoresistive microcantilever, and the biomolecules are detected [11]. Advantages of this design include small size (the cantilevers themselves may be only a few tens of micrometers in dimension), low cost, simple support electronics, and resistance to movement or vibration [10].

As the need for effective methods for the detection of biochemical entities such as virus particles becomes ever more important, nanoscale fabrication technologies are increasingly being used to create nanomechanical sensors and highly sensitive lab-on-a-
chip [11]. The applications can be very diverse and range from environmental monitoring to detecting cancerous cells to developing defenses against bioterrorism.

In one such study by Gupta, Akin, and Bashira, a nanocantilever beam is used as a mass detector, with a sensitivity of the mass of a single vaccinia virus particle. Vaccinia virus is a member of the Poxviridae family and forms the basis of the smallpox vaccine. In general, decreasing the overall dimensions of the cantilever beams results in a corresponding increase in their mass sensitivity [7].

Biomolecular sensors with the ability to ‘multiplex’, or to detect a large number of different molecular species at the same time, are being developed for serum and tissue proteomics-based cancer diagnostics, prognostics and therapeutic-efficacy monitoring [10]. Promising emerging approaches to multimolecular sensing include mechanical sensors such as the microcantilever. These comprise a large number of beams that deflect when the biomolecules of interest bind. The deflections are either observed directly by laser light or generate detectable shifts in the physical properties of the beam, such as their resonant-vibration frequency [11]. Microcantilever based, multiplexed DNA assays to detect BRCA1 mutations were recently introduced.
Typical dimensions of MEMS devices are in the several micrometers to hundreds of micrometers range. The importance of MEMS technology is not so much the size, but rather the use of planar processing technologies, related to those used in the fabrication of electronic integrated circuits, to simultaneously “machine” large numbers of relatively simple mechanical devices in an integrated fashion [14].

Nanoelectromechanical systems, or NEMS, are characterized by small dimensions, where the dimensions are relevant for the function of the devices. Critical feature sizes may be from hundreds to a few nanometers.[15] New physical properties, resulting from the small dimensions, may dominate the operation of the devices, and new fabrication approaches may be required to make them. Microelectronics fabrication technologies are driving relentlessly to manufacture smaller transistors packed with increasing density on integrated circuit chips [16]. The economic driving forces for this miniaturization are strong and have driven transistor minimum feature sizes down to the 100 nm regime. The miniaturization of commercial electronics has been taking place with an allied physics-motivated study of electron transport and magnetic properties of mesoscopic and nanoscale devices. The nanoscale studies often involve a wider range of materials and higher spatial resolution fabrication processes than the silicon microelectronics processes. Similar advanced fabrication processes can be exploited to further miniaturize electromechanical systems to bring us into the regime of NEMS [16]. The new class of NEMS devices may provide a revolution in applications such as sensors, medical diagnostics, displays, and data storage [17]. NEMS devices will enable experiments on the structure and function of individual bimolecules.[18] The initial research in science and technology related to nanomechanical systems is taking place now in a growing number of laboratories throughout the world [15].

![Fig. 3: Scanning electron micrograph of a single E. coli bacterium on an antibody-coated silicon nitride cantilever oscillator [15]](image-url)
III. Design and optimization

1. Device Structure

The structure of the device is shown in figure 4. The biosensor is comprised of a microcantilever and an external piezoelectric element for generating vibration. Piezoresistors at the root of the microcantilever are used for sensing the microcantilever’s amplitude of vibration. At the free end of the microcantilever, a small area of gold coated to capture corresponding viruses is patterned.

![Device schematic](image)

Fig. 4: Schematic of device

When the device is exposed to an environment with target viruses, the antibodies coated on the free end of the microcantilever will capture them, resulting in a resonance frequency shift. This shift is detected by the piezoresistive element. From the frequency shift, the mass of the virus can be obtained. The sensitivity of this device is defined as frequency/mass with the unit s⁻¹kg⁻¹.

2. Optimization using ANSYS

The geometry of the microcantilever is the most important parameter of the device. Different microcantilever geometries will have different resonant frequencies as well as unique frequency shifts, even under the influence of the same virus. In order to maximize frequency shift output, the geometry of the microcantilever is optimized using ANSYS. Modal analysis is used to compare the frequency shift of different geometries of microcantilevers. Among these geometries, the length, width, and thickness were kept the same, only varying shapes.

<table>
<thead>
<tr>
<th>Table 1: Chosen material properties [19]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Material properties</td>
</tr>
<tr>
<td>Value</td>
</tr>
</tbody>
</table>

Element SOLID187, a ten node tetrahedral structure element in ANSYS, was used to do the modal analysis.

The particle’s mass is simulated by using a small mass with the same density as the cantilever, with length 1µm, width 1µm and thickness 0.1µm. Making the mass about
0.285pg. The particle is located 5µm from the free end of the cantilever, as shown in Fig. 5.

![Simulating cell on cantilever](image)

**Fig. 5: Simulating cell on cantilever [19]**

We start with the basic cantilever shape shown in Fig. 6: a rectangular shaped cantilever as shown, with length \( l = 50\mu m \), width \( w = 25\mu m \).

![Rectangular cantilever](image)

**Fig. 6: Rectangular cantilever [19]**

The frequency shift is 29Hz, with natural resonance frequency 194,483Hz.

To improve the sensitivity, two parameters were considered [19].

1. Effective mass of the cantilever near the free end
2. Clamping width at the fixed end

First, we investigate how the effective mass of the cantilever near the free end affects the performance. As shown in Fig. 7 below, two cantilevers were chosen: one with increased effective mass and one with decreased effective mass. (a) \( l_1 = 20\mu m, w_1 = 30\mu m \); (b) \( l_1 = 20\mu m, w_1 = 10\mu m \);

![Shapes B and C](image)

**Fig. 7: Shapes B and C [19]**

![Shapes D and E](image)

**Fig. 8: Shapes D and E [19]**
The frequency shift of shape B is 41Hz, slightly lower than shape A, whereas the frequency shift of shape C is 69Hz. This represents a 40% increase over the frequency shift of shape A. From these results, a sub-conclusion can be made that reducing the effective mass at the end of the cantilever greatly increases the sensitivity of the structure. Second, we investigated how the clamping width at the fixed end affects the sensitivity of the cantilevers. Shape D and shape E are simulated. The frequency shift of D is 36Hz and that of E is 31Hz; the frequency shifts of both are much smaller than that of shape A. This is because the smaller clamping width at the fixed end results in a smaller spring constant for the cantilever, and thus a smaller resonant frequency.

From these investigations, we can conclude that in order to maximize frequency shift, we should reduce the mass at the free end of cantilever and increase the clamping width of the cantilever at the fixed end. Thus, it is pretty straightforward to conclude that a triangular cantilever will have optimal performance.

Therefore, a triangular cantilever, as shown in Fig. 9, is studied. The frequency shift of F is 506Hz, an order of magnitude larger than shape A. Another advantage of shape F is that the stress distribution is nearly uniform along the length of the cantilever, as shown in Fig. 10(b). Thus, there will be a greater area for piezoresistive elements. Compare this to a rectangular cantilever, wherein the stress is maximal in areas nearest to the fixed end.

Fig. 9: Shape F [19]

Fig. 10(a): Stress intensity distributions in rectangular shaped cantilever: a force 1µN applied at the end of the tip. The stress unit is MPa.
However, there is one disadvantage to using a triangular cantilever: there is insufficient area at the free end to capture the bioparticles of interest. To reduce this issue, trapezoid-like cantilevers are used in our design. Using a similar body geometry, the tip is comprised of a square with $l_1=10\text{um}$, $w_1=10\text{um}$. The frequency shift with this geometry is 162Hz. Although this is not as much as the triangular geometry, it has a greater end area for capturing bioparticles.

3. Higher resonance mode analysis

Since cantilevers have higher order resonance modes, it is worth doing a study on the frequency shift at higher order resonance. If the frequency shifts are larger, then it would be better to operate the cantilever at higher order resonance modes. ANSYS simulation was performed to investigate the trapezoid-like cantilever. With dimension $L=50\text{um}$, $W=25\text{um}$, thickness=0.5um; $l_1=10\text{um}$, $w_1=10\text{um}$.

We simulated the resonant frequencies of the cantilever with a particle and without. The results are given in table 2.
### Table 2: Shift frequency in different modes

<table>
<thead>
<tr>
<th>Order</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Resonance Frequency (without particle) (Hz)</td>
<td>264930</td>
<td>1311000</td>
<td>1970600</td>
<td>3458300</td>
<td>4616100</td>
</tr>
<tr>
<td>Resonance Frequency (with particle) (Hz)</td>
<td>264780</td>
<td>1310700</td>
<td>1970300</td>
<td>3458200</td>
<td>4616300</td>
</tr>
<tr>
<td>Shift Frequency (Hz)</td>
<td>150</td>
<td>300</td>
<td>300</td>
<td>100</td>
<td>200</td>
</tr>
</tbody>
</table>

**Fig. 12: Frequency shift in each order**

The disadvantage of using higher order modes is that higher modes have smaller vibration amplitudes, making sensing much harder. From Fig. 12, we can see that mode 2 has twice the frequency shift as mode 1, and mode 3 has the same frequency shift as mode 2. But, its vibration amplitude is smaller than mode 2, resulting in our choice of mode 2.

### 4. Sensitivity analysis

From the above, the final cantilever is as shown in Fig. 11. With dimension \( l=50\, \text{um} \), \( w=25\, \text{um} \), thickness=0.5um, \( l_1=10\, \text{um} \), \( w_1=10\, \text{um} \).

The mass of the applied particle is 0.285 pg; while the frequency shift is 300Hz (using cantilever shape G and operating at the second mode).

Thus the sensitivity is found to be:

\[
S = \frac{300\, \text{Hz}}{0.285\, \text{pg}} = 1.05 \times 10^{18} \, \text{s}^{-1} \text{kg}^{-1}
\]
IV. Fabrication

1. Process Choices

Now knowing the desired goal of an optimized cantilever, a method of achieving it through microfabrication must be decided upon. Luckily, the fabrication process of the microcantilever system proposed herein is greatly simplified through the purchase of two extant pieces: a silicon-on-insulator wafer, or “SOI” wafer, and a piezoelectric element for actuation. The use of a SOI wafer allows for simple formation of the cantilever itself using methods previously employed by groups using microcantilevers for other sensing purposes [20], other material selections were dictated by the need for biosensitivity. For this purpose, gold was selected as the binding material for the biosensitive layer at the tip of the microcantilever, as its properties are desirable [21]. However, because gold does not bind as well to silicon as it does to metals, titanium was selected as a compatibility layer to allow the gold to grow as desired [22]. This combination of metals allows the growth of a gold layer onto which a bioreactive film can be deposited through an immersion process involving placing the cantilever into a solution of the desired binding molecules and following up with a secondary bath in bovine serum albumin [23]. Because of the nature of the sensor, the desired bioreactive layer may vary from unit to unit. However, if a more complete manufacturing perspective is desired, a bioreactive layer used to illustrate the process comes from a published article regarding prostate cancer and cardiac disease [23].

2. Fabrication Process

For ease of graphical representation, the workflow diagram of the fabrication process has been broken into two primary groups. The first preparation group consists of the steps before deposition of the metals.

Fig. 13: Preparation group one: pre-metal deposition workflow diagram
The process begins with ion implantation of the SOI wafer. Due to the expense of ion implantation machinery, this step is completed by a contracted entity offsite. A mask is utilized to achieve the preferred piezoresistive geometry at the area of highest stress: the base of the unformed cantilever [24]. The dose determined to provide the optimal signal to noise ratio in the piezoresistive element was determined to be on the order of $10^{14}$ boron ions/cm$^2$ [25].

Fig. 14: Mask used for ion implantation for piezoresistive sensor (note: black sections are opaque)

Upon reception of the doped SOI wafer, the next phase of the manufacturing process can proceed. Because a photoresistive layer is required for subsequent fabrication steps, a silicon dioxide layer is grown on the upper silicon layer. The thickness of the oxide layer was determined to be of optimal thickness when it is approximately six times the total thickness of the metal films, or 950 nm. Due to the relative thickness of the oxide layer required, a wet oxidation process is employed [26]. The resultant silicon dioxide layer, after sufficient controlled exposure to ultra-violet light, is now capable of acting as a sacrificial layer for selective deposition of metals.

Following this growth of the silicon oxide layer, photolithography is employed to allow selective etching.

Fig. 15: Mask used for photolithographic step (note: black denotes opaque sections)
Exposure to ultra-violet light causes cross-linking in the silicon dioxide, resulting in a structure that is more resistant to etching than the unexposed sections. Thus, following the photolithographic step, etching the cantilever with potassium hydroxide will eliminate the oxide layer over both the tip and part of the piezoresistive element, allowing for metal deposition. However, this requires a highly anisotropic etching process to ensure that the necessary unexposed layer of oxide under the topmost exposed silicon dioxide is not destroyed. Following this process, the exposed oxide layer can be etched through exposure to sulfuric acid. This leaves one sacrificial layer of unexposed silicon dioxide.

Preparation group two begins with the selective deposition of titanium through electron-beam deposition. E-beam deposition is favored for this phase as it results in comparatively less contamination of the film and allows for careful control of the rate of deposition [27]. Additionally, the presence of the much higher silicon dioxide mask might lead to some shadowing if sputtering were to be employed, further reinforcing the choice of e-beam deposition.

Fig. 16: Preparation group two: metal deposition and freeing the cantilever

A thin layer of titanium, 5 nm is preferred [22], is desired to act as a binding layer to ensure the gold forms as desired on the cantilever tip and piezoresistive element. The
next step is the e-beam deposition of a much thicker layer of gold. The optimal thickness of this layer has been determined by other groups to be 150 nm [22]. Once the key areas have been covered in the requisite metallic film, another etching phase is needed.

This etching phase serves to eliminate the last remaining silicon dioxide on the topmost layer of the wafer. Once again, a potassium hydroxide wet etch is used to eliminate the unexposed silicon dioxide. This leaves only the titanium and gold films on the upper layer of silicon. All that remains is to free the cantilever.

The cantilever’s shape must first be defined by using deep reactive-ion etching. DRIE, being highly anisotropic in nature, is ideal for defining cuts with large aspect ratios, making it an ideal process for shaping the cantilever and freeing it [28]. Using a carbon tetrafluoride and oxygen gas mixture, the anisotropicity can be maintained by ensuring oxidative growth matches the rate of etching of the walls [29]. It is trivial to grow or deposit a photoresistive layer and expose it to protect the exposed silicon sections, and therefore it is not shown. Additionally, it should be noted that this will not affect the exposed gold layer. This then leaves the newly defined cantilever resting on top of the sandwiched silicon dioxide layer and the silicon substrate.

In order to eliminate the supporting substrate layer, bulk DRIE is employed with a similar gas mixture. This leaves only the middle layer of silicon dioxide supporting the cantilever. Once again wet etching with potassium hydroxide removes the unexposed oxide and frees the cantilever. This leaves the cantilever with a gold film on the tip and gold connecting electrode at the base for the piezoresistive element. However, the device is still incomplete.

The next step is the immersion of the device into the bioreactive bath to allow selective binding of the desired antigens to the gold film on the tip. It is of no consequence if the same bonding occurs on the gold electrode at the base, as the film is thin enough to have a negligible contribution to the overall material properties of the area. This bath and subsequent fixing baths to prevent non-specific binding allow the gold film to attract and hold proteins of interest for measurement [21, 20]. This leaves only the final step.

![Fig. 17: A diagram of cell capture following preparation of the bioreactive layer on the gold film [21]](image.jpg)

The last step in the manufacturing process is to mount the microcantilever on top of a commercial piezoelectric element.
The addition of this piezoelectric actuator allows frequency sensing for detection of the mass of proteins or cells of interest attached to the bioreactive layer by inducing oscillation in the cantilever. This provides us with a finished, optimized cantilever.

V. Conclusion

This concludes our presentation of a microcantilever-based detection device for biosensing. Herein, we have discussed the design, optimization, sensitivity analysis, and microfabrication processes involved in bringing our device into reality. By using a piezoresistor for sensing, utilization of a complex optical readout system is avoided. Based on ANSYS simulation, optimized geometry and higher mode performance are has provided a method by which the sensitivity of this device can be increased to $1.05 \times 10^{18} \text{s}^{-1}\text{kg}^{-1}$. Use of a SOI wafer greatly reduces the complexity of the process and improves its reliability. The overall goal of this device is to become a portable platform for biodetection, allowing detecting various bioparticles corresponding to the specific bioreactive films coated on gold layer.
Acknowledgements

We thank Prof. Horacio Espinosa for his insightful instructions on this project. He suggested us using external piezoelectric element instead of putting piezoelectric layer on top of the cantilever, which dramatically reduces the complexity of fabrication process.

References

**Biography**

Huan Hu (Alan) is now pursuing a Ph.D degree in Mechanical Engineering department. He got his Bachelor and Master degree in Tsinghua Univ. in China. Having been doing research in MEMS research for 3 years, he believes that MEMS have the potential to change the world. He wants to be a professor in MEMS area and contribute to the development of MEMS academy.

Ben Murphy is currently an undergraduate at Northwestern University’s McCormick School of Engineering. Currently pursuing a bachelor’s degree in Mechanical Engineering, Ben hopes to matriculate into the Master’s Program in the Mechanical Engineering department. His ultimate career goal is currently undecided, but he has a strong desire to pursue a Ph.D. and perhaps become a professor in the area of mechatronics.

Sylvester Ogletree is a senior in the McCormick School of Engineering and Applied Sciences. He is currently majoring in mechanical engineering. He enjoys reading, running, playing basketball. His ultimate career goal is to own his own business.

**Delegation of tasks**

Literature Review – Sylvester  
Design, optimization and sensitivity analysis– Alan  
Microfabrication and mask design -- Ben
Appendix

ANSYS Code ___ Mode Analysis of Shape G cantilever
(Just paste it on the command window in ANSYS, you can see it work)

/PREP7
!##----choose element Solid187
ET,1, SOLID187

!##----material characteristics
MPTEMP,,,,,,,
MPTEMP,1,0
MPDATA,DENS,1,,2.85e-15
MPTEMP,,,,,,,
MPTEMP,1,0
MPDATA,EX,1,,1e5
MPDATA,PRXY,1,,0.24

!##----create cantilever model using keypoints
k,1, 0, 12.5, 0
k,2, 40, 5, 0
k,3, 50, 5, 0
k,4, 50, -5, 0
k,5, 40, -5, 0
k,6, 0, -12.5, 0
k,7, 0, 12.5, 0.5
k,8, 40, 5, 0.5
k,9, 50, 5, 0.5
k,10, 50, -5, 0.5
k,11, 40, -5, 0.5
k,12, 0, -12.5, 0.5
K,13, 44.5, 0.5, 0.5
k,14, 45.5, 0.5, 0.5
k,15, 45.5, -0.5, 0.5
k,16, 44.5, -0.5, 0.5
K,17, 44.5, 0.5, 0.6
k,18, 45.5, 0.5, 0.6
k,19, 45.5, -0.5, 0.6
k,20, 44.5, -0.5, 0.6
v,1,2,5,6,7,8,11,12
v,2,3,4,5,8,9,10,11
v,13,14,15,16,17,18,19,20

vglue,1,2
vglue,2,3
### Create mesh

SMRT,9
CM_, Y, VOLU
VSEL,, , , 3
CM_, Y1, VOLU
CHKMSH,'VOLU'
CMSEL,S_, Y
!* VMESH_, Y1
!* CMDELE_, Y
CMDELE_, Y1
CMDELE_, Y2
!* ! vplot
CM_, Y, VOLU
VSEL,, , , 4
CM_, Y1, VOLU
CHKMSH,'VOLU'
CMSEL,S_, Y
!* VMESH_, Y1
!* CMDELE_, Y
CMDELE_, Y1
CMDELE_, Y2
!* ! vplot
MSHAPE,0,3D
MSHKEY,1
!* MSHAPE,1,3D
MSHKEY,0
!* CM_, Y, VOLU
VSEL,, , , 1
CM_, Y1, VOLU
CHKMSH,'VOLU'
CMSEL,S_, Y
!* VMESH_, Y1
!* CMDELE_, Y
CMDELE_, Y1
CMDELE_, Y2
!##----get the 5 modes of frequency as well as the animation
FLST,2,1,5,ORDE,1
FITEM,2,5
!* 
/GO
DA,P51X,ALL,0
FINISH
/POST1
FINISH
/SOL
!* 
ANTYPE,2
!* 
MSAVE,0
!* 
MODOPT,LANB,5
EQSLV,SPAR
MXPAND,5, , ,1
LUMPM,0
PSTRES,0
!* 
MODOPT,LANB,5,0, , ,OFF
/solve
solve
FINISH
/POST1
SET,LIST
!* 
! PLDI, ,
ANMODE,10,0.5, ,0
!* 
SET,FIRST
!* 
! PLDI, ,
ANMODE,10,0.5, ,0
!* 
! /ANFILE,SAVE,'Model1','avi','H:\Cantilever simulation'\
SET,NEXT
!*
! PLDI, ,
ANMODE,10,0.5 ,0
!*
! /REPEAT,RESIZE
! /REPEAT,RESIZE
! /ANFILE,SAVE,'Mode2','avi','H:\Cantilever simulation\'
SET,NEXT
!*
! PLDI, ,
ANMODE,10,0.5 ,0
!*
! /ANFILE,SAVE,'Mode3','avi','H:\Cantilever simulation\'
SET,NEXT
!*
! PLDI, ,
ANMODE,10,0.5 ,0
!*
! /ANFILE,SAVE,'Mode4','avi','H:\Cantilever simulation\'
SET,NEXT
!*
! PLDI, ,
ANMODE,10,0.5 ,0
!*
! /ANFILE,SAVE,'Mode5','avi','H:\Cantilever simulation\'
! /REPEAT,RESIZE
! LGWRITE,'Final','lgw','H:\CANTIL~1\',COMMENT