Optical sensors for high sensitivity motion detection

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NANYANG
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**OPTICAL SENSORS FOR HIGH SENSITIVITY
MOTION DETECTION**

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2016

OPTICAL SENSORS FOR HIGH SENSITIVITY MOTION DETECTION

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ABSTRACT

Optical methods providing fast detection, electromagnetic resilience and high sensitivity are discussed. In this thesis, two different optical device design concepts are discussed. One design concept employs the use of an optical gratings based accelerometer design concept that was developed for seismic motion detection purposes that provide miniaturization, high manufacturability, low costs and high sensitivity. Finite element simulations and theoretical modeling have been carried out for the optimization of design variables. Detailed in-house fabrication procedures of a double-sided deep reactive ion etching (DRIE) on a silicon-on-insulator (SOI) wafer for the MOEMS device are presented and discussed. Experimental results obtained show that the conceptual device successfully captured motion similar to commercial accelerometer with average sensitivity of 13.6 mV/G with highest recorded sensitivity of 44.1mV/G. This is the first MOEMS accelerometer developed using double-sided DRIE on SOI wafer for the application of seismic motion detection, being a breakthrough technology platform to open up options for lower cost MOEMS devices.

The second design concept involves the development of a proof of concept for a wearable cuffless blood pressure sensor. Three different approaches have been simulated for this device, with the first one employing a curved PMMA waveguide. The other two approaches employ the use of a ball shaped side-fire optical tip as a waveguide for detection of deflections along the skin due to blood pressure. Experimental results have shown that the PMMA waveguide was able to detect pulse rates when only one device is used and pulse wave velocity readings with the addition of a separate pulse transducer. This is the first optical fiber device used for the development of a wearable blood pressure sensor, and also the first application of an optical ball shaped side-fire optical fiber tip in a biomedical device.

The development of these two design concepts would open up avenues for optical devices in their respective fields. This would thus act as new platforms for optical motion sensing and optical wearables technology, bringing these fields closer to users in the long run.

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CHAPTER 1: INTRODUCTION

1.1 BACKGROUND

With repeated advancements in technology, optical engineering has been promoted in different arenas of life. In the past decade, various researchers have been furthering the progress of this field and today, optical engineering can be found almost everywhere from telecommunications industries to other optical applications like CD drives. In the aspect of security and defense industries, there is a strong reliance on information gathering and analysis to monitor and develop an understanding of the current environment. One of the means to carry this out is through placing sensors around areas of interest that would transmit data to a server that would decipher these sets of data to generate an understanding of the environment to end users (See Figure 1). This has placed a strong emphasis on the sensitivity of the sensors and devices deployed. By having sensors that are highly sensitive, it would allow end users to paint a better picture of the events that are occurring in the areas of interest, and computers would then be more able to differentiate motions and filter them according to relevance. In example, higher sensitivity of these devices would then allow computers to recognize whether it is a human or an animal that is trespassing the area of interest.

Currently there are six main modes of motion detection, namely infrared, visual imaging, acoustic, magnetic, electrostatic and seismic. Each of these methods are not mutually exclusive and are usually used as a combination to allow a more comprehensive picture, however there are some advantages to using seismic detection. Seismic motion detection is passive and has non-line-of-sight capabilities while having low power consumption and also displays wireless capabilities. This allows it to also be used as an unattended ground sensor that can be used in various terrains and situations.

Figure 1. Security uses for motion detection[1]

	Seismic	Magnetic	Active IR	Passive IR	Acoustic	Break-wire	RF Communications
Fog	-	-	◇	◇	○	-	-
Heavy Fog	-	-	●	●	◇	-	-
Rain	◇	-	◇	◇	◇	-	-
Heavy Rain	●	-	●	●	●	-	◇
Light Wind	◇	-	-	○	◇	-	-
Gusting Wind	●	-	◇	◇	●	○	-
Strong Wind	●	-	◇	●	●	-	-
Snow	-	-	◇	◇	○	-	-
Heavy Snow	-	-	●	●	◇	-	-
Deep Snow	◇	-	●	●	●	●	◇ / ●
Ice	◇	-	◇ / ●	◇ / ●	●	-	-
Dust	-	-	◇	◇	◇	-	-
Sandstorm	○	-	●	●	●	○	◇
Thunder	◇	-	-	-	◇	-	-

- Little or no Effect ○ Minimal Effect ◇ Moderate Effect ● Severe Effect

Figure 2. Relative effects of weather conditions on common sensor modes[2]

However, the seismic sensor also has its own set of disadvantages like every other sensor. Highly sensitive seismic sensors would be relatively susceptible to environmental factors as mentioned in Figure 2. It is highly dependent on the soil condition of the areas of interest and is also not able to distinguish the source of vibrations by itself and would require additional analysis software to accomplish this. Multiple sets of sensors would also need to be deployed to allow maximum coverage and also to determine the location of the source of vibrations. Although these effects are prominent, with prior calibration and software modulation, effects of such environmental influences can be greatly reduced.

Figure 3. Evolution of seismic motion detection [3-14]

Another application of such a device would be in the field of earthquake studies and detections, where early detection would help to map out real time seismic activities and predict earthquakes even before they happen, allowing early evacuation of active sites and reducing fatalities. A more recent use of seismic detectors are in the field of wildlife protection, for detecting poachers, and also immigration control, for detecting illegal immigrants sneaking through national borders. Looking at such vast possibilities for future applications, it is therefore a very interesting area to research into.

The first portable geophone was invented by Frank Alvord Perret for use by the British to detect tunneling activity within 50 meters of deployment.[15] Following the evolution of seismic motion detection (See Figure 3), there was a shift in direction for the development of seismic sensors towards miniaturization while also emphasizing on sensitivity. Moving closer to recent times, it is found that MEMS-based accelerometers are now more able to detect static gravity at microgravity-scale resolution. Seismic events and motion can thus be detected using high sensitivity MEMS-based accelerometers. Various methods of detection have been used to produce accelerometers for displacement measurement, e.g. capacitive, piezoelectric, piezoresistive, optical, etc. Making use of current MEMS manufacturing infrastructure

Figure 4. Optical approaches in the biomedical industry[16-20]

provides great benefits of miniaturization, lower cost, mass production of highly precise devices with low tolerances, and also allows different techniques to be applied in the packaging of the chip that prevents random vibration of air molecules, reducing instrument noise and temperature insensitivity.[21, 22] Although these current devices have proved to be successful and have high sensitivities, they are still relatively susceptible to electromagnetic interference and harsh environments when not enclosed in bulky packaging, rendering them unusable in certain situations. By employing a relatively novel approach in this field, which is the optical approach, it is possible to overcome such difficulties and also yet providing sensitivities that are on par with current designs. Studies have also shown that the frequency range of human seismic motions from walking would typically fall within the range of 0 to 500Hz.[23-26] Bard et al [27] also estimated that the vertical force exerted from a human footstep would be estimated to be around 80% to 120% of his body weight.

Another field where optical motion devices are greatly advantageous would be in the biomedical field (See Figure 4). This is due to the high demand for sensitivity and accuracy. Blood pressure measurement enables an individual to obtain arterial blood pressure measurements which is vital in evaluation of one's cardiovascular system for diseases like hypertension, atherosclerosis, heart failure, etc.[28] Traditional methods comprise of clinicians using sphygmomanometers with cuffs and stethoscopes for hearing Korotkoff sounds that determine the diastolic blood pressure (DBP) and systolic blood pressure (SBP) of individuals. In specifics, the medical industry has been greatly interested in developing a device that would enable individuals to measure their blood pressures outside a clinic as there are many

restrictions within the clinic itself that would prevent an accurate assessment of an individual's physical condition.

Some advantages of wearable BP sensors are that short term fluctuations in blood pressure can be observed across a day due to the circadian rhythm of an individual, use of various drugs and caffeinated products, type of activities, psychological state eg. white coat effect, stress, drowsiness, etc.[29] It has also been recorded that systolic BP varies approximately 10-15 mmHg in the day and 5-10 mmHg at night, and that caffeine causes 5-10 mmHg increase in systolic BP too.[30] These differences in measurements could give a wrong indication of an individual's health condition, which could result in unnecessary scares to users, however these can be easily overcome through programming algorithms which would not be discussed in this thesis. Apart from these areas of variability, wearable BP measurement devices have to accurately detect and display measurements given the non-ideal, yet realistic conditions the user is in. The medical industry has continually improved their technology, however wearable technology for this means are currently still limited to cuffed ones. Cuffed blood pressure monitors can cause unnecessary irritation during wear, therefore there is a demand for cuffless measurement. Also, cuffless measurement would allow greater opportunity for integration with wearables. Optical devices are also more useful in this aspect as they have to be operational in places with high electromagnetic influences and on top of these, optical fiber sensors would be more flexible for use as biomedical devices.

1.2 OBJECTIVES

This project's main aim is to demonstrate a series of prototypes that would be able to push the boundaries of optical sensing and motion detection in terms of accuracy and sensitivity. In this project, two types of design prototypes are developed. The first is an optical Micro-Electro-Mechanical-System (MEMS) device that would be used for low acceleration (low G) applications. The novelty in this would firstly be in its design, that has been optimized based

on the natural frequency and displacement, and also its novel application of the complex double-sided deep reactive ion etch (DRIE) on silicon-on-insulator (SOI) for the optical MEMS layer. The observable output of the device is targeted to display a minimum of 0.1 microns displacement upon an external force of 2G (where $G=9.81\text{m/s}^2$) with resonant frequency at approximately 2000Hz. With the measured resonant frequency at this number, the range of frequencies that would give a linear response would be approximately from DC to 500Hz, which is 25% of the resonant frequency given. The direct implementation would be in the field of highly sensitive seismic human motion detection by making use of optical grating method of detection. For this prototype, it is targeted for the device to be able to detect seismic forces when a human steps on the ground right beside it. In this scenario, the force on the ground would then be as close to the force experienced by the mass.

The second is a wearable cuffless blood pressure measurement sensor that would accurately detect a user's blood pressure through the use of the optical fiber or optical waveguide approach. The first novelty of this device would be in its approach as an optical device for cuffless wearable blood pressure measurement. The second novelty in this device would be the integration of a new optical fiber tip that enables side-firing capabilities to illuminate a part of the skin where slight displacements due to blood pressure can be detected. Apart from this, this device would also be advancing the technology for blood pressure sensing as being an optical cuffless wearable device. Through the development of these new device concepts, new platforms for optical blood pressure detection will be developed.

1.3 SCOPE

The scope of this project is to firstly develop a reliable design concept for the optical MOEMS sensor followed by fabrication of the MOEMS layer. Theoretical and finite element simulations are to be done to study relationships between design parameters and are optimized according to project objectives. Fabrication is then carried out and testing to be done on prototype. The secondary portion of the project is to develop design concepts for a novel wearable cuffless blood pressure measurement sensor which would be able to effectively provide blood pressure measurement values with high accuracy.

CHAPTER 2: LITERATURE REVIEW

2.1 SEISMIC MOEMS DEVICE

2.1.1 MECHANICS OF FOOTSTEP DETECTION

When people or vehicles move about over ground, a series of impacts are experienced over the ground and are then directed away from the source by various forms of seismic wave propagation, which enables detection through highly sensitive devices.[31] Even though there is high exposure to environmental factors that would generate noise, various techniques have been developed to either eliminate the detection noise through closed loop feedback or to reduce its significance through normalization of results using kurtosis and cadence values. Figure 5 below shows the breakdown of different forms of wave propagation when there is an impact delivered unto the ground. Upon impact, different modes of vibration are generated in the ground and are attenuated in different directions. The shear and compressional body waves (S and P waves) are propagated into the solid ground, expanding in a spherical direction unto the earth's core [32] which make it harder for detection. Love waves are horizontally polarized shear waves that usually travel faster than Rayleigh waves and have the highest amplitude, but depend on soil property gradients, making it hard for general application. Surface waves like Rayleigh waves travel along the Earth's surface with longer duration and larger amplitude are the main forms of seismic motion that are currently used for detection of seismic activity. In this case, this form of wave propagation is the most suitable wave used for detection of motion. For vehicle motion detection, Rayleigh waves are also the main waves of concern. In comparison to human footsteps, the main differences would be in the frequency and amplitude of impact.

Figure 5. Diagram on wave propagation[32]

2.1.2 HUMAN MOTION DETECTION

There are various ways to detect and track human motion, and they are categorized into infrared, acoustic, seismic, optical imaging, electrostatic and magnetic detection. It is understood that there is not one method that is the best amongst them, but rather, each method serves a different purpose and application. It is also possible for two or more methods to be used concurrently for greater characterization of motion. However, with respect to defense and security scenarios, the various advantages and disadvantages will be discussed.

An infrared motion sensor usually consists of an optical element and an infrared (IR) detector that collects radiation from a certain viewing range.[33] Infrared light will be emitted from the optical element unto objects within the viewing range, while the detector produces electrical signals that are relative to changes in the surface blackbody temperature of bodies within the viewing range. A digital signal is sent from the analog processing circuit which is compared with a certain threshold level for temperature change to determine if there is any motion. IR sensors are advantageous as they are inexpensive and have fast speed response, but

on the other hand, it does not effectively confirm if the detected heat source is due to a person passing through the viewing field. It is also relatively insensitive to mechanical and acoustic noise.[34]

Acoustic detection makes use of piezoelectric microphones to detect sounds produced during motion.[24] Some good points of this detection method is that it is low cost, compact, and with low electric noise.[23] But it is highly affected by environmental effects, like the presence of noise generating machinery, and have a limited field of detection especially in open spaces.[35] There is also difficulty in distinguishing human and animal motion using a single ultrasonic sensor.

Optical imaging makes use of infrared or visual imagers to detect motion. Videos are captured in the process of detection for human gait detection and object recognition. Infrared imagers are similar to infrared sensors except that they make use of imaging techniques and can also be used to observe motion of any other kinds of hot bodies and surfaces. This method of motion detection is currently very popular in the research field and has very wide range of implementation principles.[36, 37] Highly accurate and precise results can be achieved using optical imaging, but at the same time, these devices are costly to implement and require certain amounts of knowledge and experience for practical usage.

Magnetic sensors are mainly used to detect ferromagnetic materials carried around by people, like during security checkpoints, and also in localized detection for biological studies in humans. Electrostatic sensors detect the built-up electrical charge on people, and are also rarely used for human detection across a large field. These two methods are not suitable for large scale footsteps tracking and motion detection in humans.

Seismic detection method employs the use of vibrations that are resultant of a person's footsteps during motion as mentioned in the introduction. By detection of these vibrations, low-frequency seismic energy can be detected by sensors placed within vicinity. Some

disadvantages for this would be that multiple sensors are needed for accuracy in pinpointing the position of humans, and that the attenuation of vibrations would be different for different surface types (rock, soil, tiled floors, etc.).[38] Seismic detection is also unable to explicitly define the motion of a human. Although this, it has a wide coverage in terms of field of detection as compared to all the other methods and is also relatively low cost.[39] Seismic detection is also considered relatively covert as it is passive, does not require any projection of signals unto human subjects and is relatively less influenced by environmental conditions (See Figure 2).

2.1.3 SEISMIC MOTION DETECTORS

There are different kinds of devices used for seismic human detection currently, namely the geophones and seismic accelerometers. The basic mechanism of a geophone is made a cylindrical coil suspended in a magnetic field provided by a small permanent magnet. By doing so, an oscillatory system is created with a resonant frequency dependent on the mass of the spring and the stiffness of the suspension. The geophone is then implanted into the ground with a spike attached to the base of the external casing for good ground coupling (See Figure 6).[40] The geophone is only useful when it is used for detection above its resonant frequency as it is where the output is linearly proportional to velocity.[41] Also, there is a need for accurate phase correction for accurate travel time and amplitudes, which is quite complicated. The main advantages of geophones are that they are able to operate without any electrical power, lightweight, robust and are able to detect extremely small ground displacements.[42] However, research studies are currently looking at replacements of these analog electromagnetic coiled geophones through accelerometers.[43] In this way, using MEMS accelerometers creates the possibility of reducing costs while improving data quality, however it requires the use of multiple accelerometers to reduce ambient noise.[44]

Figure 6. (Top) Field geophone with spike. (Bottom) Exploded view of a geophone[40]

Accelerometers used in this field are often piezoelectric. Some of the advantages of this piezoelectric approach is that they are completely self-contained and self-generating from an inertial frame of reference, have high sensitivity with low non-linearity, and also low power dissipation.[45] These properties have also attracted many researchers into further advancing this technology, but it does not have the same advantages as the optical approach as it cannot be used in electromagnetic and harsh environments unless proper packaging with sufficient electromagnetic shielding is designed for it. A table of accelerometers developed for seismic detection is given below with their specifications and it can be observed that most of those devices used are piezoelectric (See APPENDIX A). Also, one main disadvantage for usage of optical/laser devices would be the small relative vibrations experienced by the optical

equipment, contributing to the system's noise level. This is because the optical method is very highly sensitive.

2.1.4 SILICON MATERIAL

To fabricate a device, one major component would be the material used for the fabrication of the product. Current technology mainly makes use of silicon wafers as substrate material for the development of electronic devices. Silicon is one of the most commonly used materials for such devices due to its good electrical, chemical, optical and mechanical properties (See Table 1).[46] Pure Silicon is lightweight and strong with negligible initial stress due to its purity, and its mechanical properties do not change significantly with the increase of temperature. The onset of deformation only takes place at 700°C while also being inert to most chemicals and non-toxic.[46] Silicon has good electrical properties,[47] which allow it to be used as a semiconductor to make electrical circuits. It also has relatively good thermal conductivity allowing heat to be transferred effectively to target regions, and also has high melting point too. This allows the devices used to be used in relatively hot environments without breakdown. Silicon also has a high index of refraction of 3.4,[48] making it highly suitable for use as a material for making optical instruments.

An improvement on this technology is the Silicon-On-Insulator technology, whereby multilayered silicon wafers are precisely engineered to provide new functionality for more intricate and advanced silicon devices. Chips made with this new technology provide advantages due to the vertical isolation created by the buried oxide layer (BOX layer) between silicon layers (See Figure 7). The BOX layer protects the thin active layer, or also commonly known as device layer, from parasitic effects induced by the bulk layer, or also known as handle layer. Some good examples would be the leakage currents, radiation-induced currents, latch-up effects, etc.[49]

Table 1. Properties of Si and Si Alloys[50]

Mechanical Properties	
Young's Modulus (GPa)	160
Yield/Fracture Strength (GPa)	7
Poisson's Ratio	0.22
Electrical Properties	
Relative Permittivity (ϵ_r)	11.7
Dielectric Strength (V/cm $\times 10^6$)	0.3
Electron Mobility (cm ² /Vs)	1500
Band gap (eV)	1.12
Physical/Thermal Properties	
Density	2.4
Melting/Sublimation Point (°C)	1415
Specific Heat (J/g K)	0.7
Coefficient of Thermal Expansion (CTE) (10 ⁻⁶ /°C)	2.6
Thermal Conductivity (W/m K)	157

The implementation of this technology also reduces the need for complicated designs for trenches and well formations due to process limitations while also allowing more precise control of layer thicknesses when the BOX layer acts as an etch stop layer.[46] No special tools are required in the fabrication process as compared to normal silicon wafers. Although this, this technology comes with higher manufacturing costs as each SOI wafer is known to be more expensive compared to the normal silicon wafers. There is also a cause for concern regarding the differential stress in the thin device layer, which could reduce efficiency of certain devices. To go around the issue of costs, the number of devices fabricated on a single

Figure 7. An example of a device fabricated using SOI wafer[51]

SOI wafer is maximized as much as possible. With regards to the differential stresses, as long as this is not an important consideration in the design of the device, it would be fine.

2.1.5 MOEMS

MEMS devices that make use of optics, in short, MOEMS, are miniature systems that combine optics, micromechanics and microelectronics in accordance to its type and field of application. These devices require precise interfaces and integrated components to have reliable and highly accurate results. They can also be used in harsh environments with highly explosive atmospheres or under strong electromagnetic fields,[52] and also have high sensitivity and stability with relatively low noise levels.[22, 53, 54] MOEMS devices are also commonly found in different industries, ranging from biomedical to space exploration and also environmental monitoring, with specific applications in scanning and sensing devices, projection and display, and also gyroscopes and accelerometers. Under this huge umbrella of optical methods, there are many different ways to measure acceleration.

Figure 8. Comparison of similar MOEMS motion detection devices with sensitivity in $\mu\text{m/G}$

Various devices developed using this MOEMS method for motion detection has been studied and a list of them are given in APPENDIX B, with critical specifications mentioned. One crucial criteria of the devices is the sensitivity of the devices, which have various units. Figure 8 shows a summarized chart of devices with different sensitivity. With the various optical methods, Table 2 shows the advantages and disadvantages of each method used for these devices. Among all these detection methods, it is observed that the least complicated and most straightforward method would be to implement the method of optical detection using gratings for the developed device. This method involves the direct flooding of light on gratings that are directly etched on the proof mass of each device. With this, the phototransistors placed below the proof mass would be able to detect the intensity of the incident light and thus providing an output reading that is proportional to the magnitude of input force. These few devices only involve direct measurement of light upon the phototransistors without any further implementations. Also, they would allow better quality results as compared to the other

Table 2. List of devices developed based on different optical methods

Author	Sensitivity	Resolution	Resonant Freq	Noise	Method
Abbaspour-Sani, E.[73]	40 mV/G	-	3.2kHz	40 mV	Optical detection using gratings
Keeler, B. E. N.[22]	-	160 fm/ $\sqrt{\text{Hz}}$	6 to 80 MHz range depending on beam width	-	Optical detection using gratings
Olsson, R.H. [69]	-	50 fm/ $\sqrt{\text{Hz}}$ @ 10mHz	-	-	Optical detection using gratings
Krishnamoorthy, U. [51]	5900V/G or 20 $\mu\text{m}/\text{G}$	12fm/ $\sqrt{\text{Hz}}$	1131Hz for gimbal	8 nG/ $\sqrt{\text{Hz}}$	Optical detection using gratings
Hortschitz, W.[70][71][72]	60nA/nm	8pm/ $\sqrt{\text{Hz}}$	1700Hz	80 nV/ $\sqrt{\text{Hz}}$	Optical detection using gratings
Yao Ge[53]	300 mV/g up to 60 Hz	-	135 Hz	-	Optical Fiber
Guan, T.[54]	1.5 mV/g	-	1000 Hz	-	Optical Fiber
Llobera, A. [52]	34.3 $\mu\text{m}/\text{g}$	-	85 Hz	-	Optical Fiber
Doyle, C.[55]	1.1 $\mu\text{m}/\text{G}$	-	592 Hz	-	Optical Fiber
Conforti, G.[56]	10 mV/ μm	0.1 μm at 100 Hz	100 Hz	-	Optical Fiber
Zuwei, Z.[57]	130 kHz/G	-	39.1 MHz	-	Optical Fiber
Schröpfer, G. [58]	1.8 V/g or 143 nm/g	-	1176 Hz	-	Optical Fiber
Zandi, K. [59]	2.5nm/g	400 μg	1670Hz	-	Optical Fiber
da Costa Antunes, P. F.[61]	11 nm/G	-	38.836 Hz	-	Optical Fiber
Borinski, J. W.[60]	40.8 nm/G	0.025 g minimum detectable level	2000 Hz	-	Optical Fiber
Wu, Y.[62]	654.7 mV/G	-	146 Hz	-	Optical fiber Loop
Su, W.[63]	2 V/g	2.5 mg	3500Hz	5mV	Photoelastic effect
Tang, D.[64]	29.4 mV/G	-	4000 Hz	-	Photoelastic Effect
Loh, N. C.[65]	39 $\mu\text{m}/\text{G}$ or 283 mÅ/ $\sqrt{\text{Hz}}$ at V/mG	10 Hz	80 Hz	-	Interferometry
Cooper, E. B.[66]	-	2 mg/ $\sqrt{\text{Hz}}$	906 Hz	90ng/ $\sqrt{\text{Hz}}$	Interferometry
Andò, B.[67][68]	6.76 %/(m/s)	-	-	-	Photonic Band Gap Detection

methods which would require more complicated fabrication procedures and setups, which can cause higher possibilities of error. One main consideration for this method of detection would be the internal reflection of light that it will experience during operation which can be observed using optical simulation software, however, if the effects are properly studied, this can be neglected. A series of reviews of each device employing this method of detection is given below.

The optical accelerometer developed by Abbaspour-Sani et al [55] in 1995 made use of two different layers of n-type silicon wafers stacked up upon one another. One layer was used as a shutter and the other was used as a substrate where PIN diodes were fabricated on to act as photodetectors. On the shutter layer, a seismic mass was etched with gratings to allow light to propagate unto the detector layer from a light source placed above the shutter (See Figure 9) for measurement of PIN diode current. The seismic mass was attached to the whole shutter layer using two beams, allowing a parallel displacement instead of an angled motion under an applied force. (See Figure 10) There also is a reference diode area beside the seismic mass detection area that is operating under dark conditions which is meant to compensate for the temperature drift of the photodetector. Experimental results have shown a linear static device response in the range of $\pm 84g$, and a flat relative sensitivity for dynamic response of up to 700Hz.

Abbaspour-Sani's device could be considered as one of the earlier devices developed using this method, thus explaining the relatively lower range of sensitivity as compared to the others. Although this, the ideas of using a reference diode to cancel out the effects of temperature drift and also coating a layer of silicon dioxide to prevent internal reflection are quite innovative. The total number of layers for his device was also minimal, reducing integration and possible misalignment errors. In his paper, he also tested using a single beam shutter as it would have produced a larger displacement effect upon external force, which

unfortunately experienced rotational motion. Thus that idea was given up and the two beam device was used. Although this, there is still no consideration of the cross axis sensitivity of the device as the proof mass could still deflect in the out-of-plane direction, rendering the output inaccurate. Still, the overall concept was successfully demonstrated.

Figure 9. Device design: (a) LED light source (b) MEMS shutter (c) photodiode layer

[55]

Figure 10. (a,b) Dual beam shutter before and after applied force (c) Bending of single beam shutter under applied force [55]

Keeler et al [22] has managed to produce a highly sensitive device using the same method of detection, but on a nano-scale level. This makes it an ideal transducer element that allows detection of pico-scale motion. The nanostructures were fabricated using amorphous diamond in a surface micromachining process, which makes the device very costly to produce. Keeler's approach was subsequently further improved and integrated into the device mentioned in Olsen et al [56] which had a measured sensitivity of $50 \text{ fm}/\sqrt{\text{Hz}}$ @ 10mHz. Although this device requires the use of nanotechnology manufacturing, which is of a much more complex method of fabrication as compared to MEMS technology, nonetheless, it can still act as a good benchmark for this project due to its high sensitivity.

In another more recent study, Krishnamoorthy et al [53] developed another accelerometer making use of nano-gratings have managed to push their boundaries by having their device resonant frequency as low as 36Hz with thermal noise floor as low as $8\text{nG}/\sqrt{\text{Hz}}$. The device makes use of a proof mass attached to a movable gimbal, which results in a multiple mode of freedom approach as compared to the traditional fixed frame approach. This makes theoretical analysis more complex, but unfortunately it was not described in the paper itself. With regards to the fabrication process, the device is fabricated using surface micromachining with DRIE on both sides of the wafer, which means that alignment issues may arise. High specificity also required the device to have special procedures during the fabrication process, making it a tricky issue to overcome.

Hortschitz et al [57, 58] has also presented a hybrid MOEMS displacement sensor in 2011 and 2012. This device made use of an external LED source illuminating a proof mass and photoresistor below (See Figure 11), but it comprised of 2 sets of gratings instead. A layer of chromium is patterned onto a glass chip before being bonded to the silicon proof mass with another set of gratings. With one set of fixed chromium patterns and another movable silicon mass grating, the amount of light passing through to the phototransistor is filtered in proportion

to the displacement of the device. Initially, there were two different designs for the chromium pattern (See Figure 12), but only one was used eventually as it resulted in less scattered light flux through the aperture upon experimentation. When a force is experienced, the MEMS stencil would have a displacement in the direction of the arrows in Figure 13, changing the area of the holes where the light would pass through, which would in turn change the photocurrent generated in the phototransistors below. Experimental results and theoretical calculations have also proved that by lowering the resonance frequency, the displacement resolution for low-frequency vibrations would thus increase. Experiments were carried out using a proof of concept and results obtained revealed a high sensitivity of 0.85mV/nm for the displacement of the mass, and a corresponding noise level of 14pm/ $\sqrt{\text{Hz}}$.

Figure 11. (Left) Schematic diagram of MOEMS, (Right) Micrograph of sensor structure with 1720 quadratic holes [58]

Figure 12. Illustration of two Chromium patterns (a) Opaque squares in Cr pattern, (b) Square holes on Cr pattern [57]

Figure 13. Schematic drawing of mechanical system of MEMS sensor with variables

[57]

In Hortschitz's paper, there was no mention of internal reflection in between the different layers and it was assumed that there were no misalignment errors during the bonding of each layer unto one another, which are crucial factors that would affect the overall results if not considered. In this device, it is also observed that there are a total of 1720 quadratic holes. Although this amount of holes would help to increase the amount of light reaching the phototransistors, it also meant that any small insignificant error (due to fabrication, misalignment, etc.) would also be amplified, increasing the error percentage of each device itself. Overall, this device is a very innovative design, but if it were to be mass produced, there would also be low yield due to the inaccuracy of the device albeit the high sensitivity.

With these information, a design concept based on the spring-mass-dashpot theory would be developed, taking the advantages of the devices developed by Abbaspour-Sani and Hortschitz, allowing large bandwidth, minimizing cross axis sensitivity and also having reduced misalignment errors. The device designs would be mentioned further together with the theory in the next chapter, with optimization of important variables carried out using simulations also given.

2.2 WEARABLE OPTICAL BLOOD PRESSURE SENSOR

2.2.1 MECHANICS OF BLOOD FLOW

Figure 14. Schematic diagram of blood vessel wall [59]

Blood pressure measurement is a measurement of the force experienced by the walls of the arteries as blood is pumped from the heart throughout the body. The basic schematic of the muscles in a blood vessel is given in Figure 14, where the thickness of each layer of muscles differ according to the different types of vessels and locations in the body. Although this, the pumping motion generated from the opening and closing of the heart valves can still be observed from the waveforms captured by measurement devices placed near different vessels that are near the skin surface. A series of information can be gathered from a measurement, namely the systolic blood pressure (SBP), diastolic blood pressure (DBP), mean arterial pressure (MAP), pulse pressure, pulse rate, pulse arrival timing and pulse wave velocity (PWV). It is also found that the SBP increases as the distance travelled for each pulse increases as in Figure 15. Across the whole body, the systolic upstroke before the peak is representative of the left ventricular ejection in the heart, with the SBP being determined by the stroke volume, velocity of left ventricular ejection and preload, systemic arterial resistance, distensibility of aortic and arterial walls, and viscosity of individual's blood. DBP is also affected by the

viscosity of blood, arterial distensibility, systemic resistance and also length of the cardiac cycle.[60] Pulse pressure is the difference between the SBP and DBP readings, and MAP is the mean perfusion pressure throughout the cardiac cycle.

Given that there are different blood pressure waveforms due to the different measurement locations across the whole body, for the assessment of health conditions like high blood pressure or cardiovascular diseases, the brachial or radial blood pressures are

Figure 15. Amplification of blood pressure waveform as it travels from the aorta to the radial artery. [61]

usually taken as they are non-invasive and are relatively easy to measure. However there has been various studies showing that there are various factors that affect the amplification of the waveform between the peripheral arteries and the central arteries differently between individuals. Also, effects of certain drugs have different effects on the amplification of the waveforms within individuals too. Therefore for a more appropriate representation of the actual blood pressure at the heart, the central blood pressure at the aortic arch would be more suitable for clinical assessment.

In order to obtain the waveform from the central artery, it would require either the invasive method of measurement or the use of estimation algorithms. Some of these methods are the General Transfer Function (GTF) method, N-point Moving Average (NPMA) method and the SBP2 method. The GTF method is a method that characterizes the relationship between

the central ascending aortic pressure with the brachial or radial pressure into a transfer function (See Figure 16). The transfer function between two sites is defined as

$$H_{(A-B)} = \frac{P_B(\omega)}{P_A(\omega)}$$

or,

$$H_{(A-B)} = M_{A-B}(\omega)e^{i\varphi_{A-B}(\omega)}$$

where $P_A(\omega)$, $P_B(\omega)$ are the blood pressure values in the frequency domain at sites A(ascending aorta) and B(peripheral artery), and $M_{A-B}(\omega) = \frac{M_A(\omega)}{M_B(\omega)}$ is the modulus of amplification (blue curve), and $\varphi_{A-B}(\omega)$ is the phase of the transfer function (red curve). This method has been tested in a previous study showing high consistency between the ascending aortic pressure with the brachial and radial pressures for low frequencies of up to 4 Hz.[62]

Figure 16. Generalized Transfer Function used in the estimation of central blood pressure from radial blood pressure, adapted from [62]

The NPMA method uses a moving point average smoothing approach to determine central pressure values, removing all pulse wave features that are above 4 kHz. However, the waveform itself cannot be produced, and drug effects cannot be accurately determined by this method too.[63] The SBP2 method is based on the observation that the pressure at the second systolic peak is identical or closely related to the central SBP, which can be quite elusive at times, especially in elderly with stiffer arteries and in younger people where the SBP2 is merged with the dichrotic notch. [64, 65] With these set of information, it is deduced that the GTF method would be a reliable, universally applicable method for use in a blood pressure device. Through the use of such estimation algorithms, it is believed that a more accurate portrayal of a person's health status will be obtained.

With respect to the skin layer (See Figure 18), much research has been conducted across the globe to determine the mechanical properties of the individual elements and layers in the skin,[66, 67] however, it would be too complex to integrate each individual element into the calculations and it is not in the scope of the project to derive a foolproof set of equations to

Figure 17. Different wavelengths of light and their penetration into the skin.[68]

Figure 18. Human skin diagram [69]

calculate the resultant displacement of the skin due to the pulses of blood flowing through the blood vessels beneath. This project thus adopts the results obtained by another researcher's findings, Laurent et al [70], for use in the theoretical proofing of this method of detection. In his paper, the author found that the maximum deflections of the radial artery walls are 0.04mm in diameter across the cross section of the artery itself. Using this, it is deduced that the estimated deflection tangential to the artery walls would be approximately 0.02mm, assuming

an equal radial expansion of the artery during pulsation. In addition to this, the artery walls are directly in contact with the dermis layer in the skin, therefore an expansion of the artery wall would produce a corresponding force unto the skin layer, which in turn can be detected through devices placed on the skin.

2.2.2 EFFECTS OF LIGHT ON SKIN

In 2012, the Scientific Committee on Emerging and Newly Identified Health Risks (SCENHIR) concluded an opinion on Light Sensitivity that identified blue light and ultraviolet radiation as a potential “risk factor” for aggravation of light sensitive symptoms in certain patients.[68] The UV band is divided into three wavelength regions (CIE 2006/62471):

- UVA from 400nm - 315nm
- UVB from 315 nm - 280nm
- UVC from 280 nm - 100nm

And the IR band into three wavelength regions too (CIE 2006/62471):

- IRA from 780 nm - 1400nm
- IRB from 1400 nm - 3000nm
- IRC from 3000 nm - 1mm

Studies have shown that the light interacts with the skin layer in all four different ways, transmission, reflection, absorption and scattering, passing through the skin as according to Figure 17. These interactions are determined according to the spectrum of light introduced and also the constituents of the irradiated skin area itself. As this topic can be very deep, the effects of light on the skin in the cellular level would not be touched on, but rather a brief introduction of the more observable changes would be discussed. Absorption of light in human skin are due to chromophores that are found in blood, water, melanin, fat. Chromophores relate to the multiple bonded atom of the molecule in question which are responsible for the optical absorption properties of overall skin area.[71] A few examples of chromophores include

Figure 19. (a) Absorption coefficient and (b) scattering coefficient of in-vivo dorsal forearm obtained through observations from 18 individuals. [72]

nucleic acids, urocanic acids, proteins, melanins, etc. Their absorption spectrum are within the UV range, and different humans have different compositions of each constituent in their skin, causing different absorption and scattering coefficients in each individual (See Table 3). It was also mentioned in another paper that in the 600-800nm region, absorption coefficients decrease monotonically as wavelength increases (See Figure 19), and the main contributors to this trend are haemoglobin and melanin.[72] Therefore, the best wavelength range for use in blood pressure measurement would be in this range as it would have the most detectable fluctuation in the reflected light.

The transmission of light into the skin is mainly determined by the wavelength of light used. It has been widely reported that ultraviolet (UV) radiation causes photochemical reactions on the skin and other organic materials, which can lead to inflammatory reactions, thus it would be good to avoid having to expose the skin to such wavelengths, which is from 100nm to 400nm. A study in 2005 made use of four different visible wavelengths (463nm, 543nm, 571nm and 634nm) and have measured their penetration depth into the skin, which are approximately 0.2mm for 463nm, 0.4mm for 543nm, 0.5mm for 571nm and 0.62mm for 634nm.[73] In Figure 17, the different penetration depths of the various light wavelengths into the skin are given.

Table 3. Absorption and scattering coefficients of different human tissue types and phantoms. [74]

Tissue	Wavelength (nm)	μ_a (mm ⁻¹)	μ'_s (mm ⁻¹)
Caucasian dermis ($n = 12$)	633	0.033 ± 0.009	2.73 ± 0.54
	700	0.019 ± 0.006	2.32 ± 0.41
	900	0.013 ± 0.007	1.63 ± 0.25
Negroid dermis ($n = 5$)	633	0.241 ± 0.153	3.21 ± 2.04
	700	0.149 ± 0.088	2.68 ± 1.41
	900	0.045 ± 0.018	1.81 ± 0.040
Subdermis ($n = 12$)	633	0.013 ± 0.005	1.26 ± 0.34
	700	0.009 ± 0.003	1.21 ± 0.32
	900	0.012 ± 0.004	1.08 ± 0.27
Muscle ($n = 1$)	633	0.121	0.89
	700	0.046	0.83
	900	0.032	0.59
Phantom ($n = 7$)	633	0.027 ± 0.007	2.14 ± 0.1

Research has also found that the attenuation down to 1% occurs for light wavelengths occur at depths of 40 μ m for wavelength of 250nm – 280nm, 100 μ m for 300nm, 190 μ m for 360nm, 250 μ m for 400nm, 400 μ m for 700nm, 800 μ m for 1200nm, 400 μ m for 2000nm, 1 μ m for 2500nm, and 30 μ m for 400 μ m wavelength. It was also found in another research paper that absorption of light is minimized within 700 to 1300nm wavelength of light, which is termed the window of transmission.[75]

For the infrared (IR) region, longer wavelengths are blocked at the epidermis layer, meaning that the absorption of IR is high. This would in turn result in the heating up of the skin, which means that care must be taken to ensure that the power levels of the light source does not cause heating damage on the skin. Below 700nm, the absorption in the blood is high, which in the case of blood pressure measurement, should be avoided. Wavelengths above 2500nm are highly absorbed by water, which also should be avoided in this case.

Absorption of optical radiation on skin can also cause heating that raises the skin temperature, causing cell and tissue damage in 15 to 60 minutes at 45 degrees Celcius, and also within a few seconds at 60 to 70 degrees Celcius. UV radiation on the skin also causes formation of vitamin D3, however surplus of this vitamin causes increased risk of undesirable

effects like burning. Apart from this, inflammatory reactions happen at higher levels of UV radiation with long term effects of gene mutation of epidermis, and loss of collagen in dermis layer and photo-aging effects. It is also suspected that IR radiation may also increase skin cancer formation from sun exposure and that IRA causes skin aging. Threshold for irradiance as given in international standards DIN 33403 (DIN 2001) is given as 1kW/m^2 for exposure time above 5 min, 460 W/m^2 at 30 degrees Celcius skin surface temperature. The Commission further stated that thermal effects from visible or IR radiation from lighting sources are unlikely to cause serious health effects in healthy skin, however problems may arise with excessively intense sources with exposures in close proximity to source as according to DIN 33403 for pain thresholds or limiting to exposure times of over 10seconds.

Optical scattering has two different scattering theories which are appropriate.[76] The Mie theory considers scattering by particles in a mixture with refractive index different from its surrounding medium. Rayleigh theory considers scattering through a medium with a continuous but fluctuating refractive index where the scattering due to small particles or mass density fluctuations are much smaller as compared to the wavelength of light. These two theories form the main basis of all studies on optical transmission and have been widely studied on by various researchers. As such, it would not be touched on further in this dissertation.

2.2.3 METHODS OF NON-INVASIVE BLOOD PRESSURE MEASUREMENT (NIBP)

There are various methods of detection of blood pressure, namely the auscultatory, oscillometric, tonometry, volume clamped, pulse wave velocity methods, and their strengths and limitations are discussed below. Auscultatory method is a traditional method whereby a stethoscope and sphygmomanometer are used. The sphygmomanometer cuff is initially inflated and slowly deflated until the initial Korotkoff sound can be heard through the stethoscope, which is when the blood begins to flow through the suppressed artery underneath.

The blood pressure reading upon the initial sound would be the SBP and the point where the Korotkoff sound is completely gone would be the measurement for the DBP.[77] It is considered as the gold standard for non-invasive blood pressure measurements and is still the most widely used method for medical examinations. A limitation that it currently experience is that it is highly dependent on the user and requires medical knowledge for use. It makes use of a cuff, causing decreased comfort after long time of monitoring. [78]

The oscillometric method makes use of an electronic cuffed device that detects the oscillations observed at cardiac frequency that increases in amplitude as cuff pressure drops from SBP to MAP and then decrease as it falls further from MAP to DBP. BP measurements are then interpreted using a computer or processor attached.[79] This method is currently the most widely used method for long term monitoring and the automatic measurement does not require any skilled operator. Likewise with the auscultatory method, it requires the use of a cuff and also decreases patient comfort for long time monitoring. On top of that, it is also sensitive to stiffness of arteries and also highly dependent on the measurement site. [78]

Arterial tonometry is a continuous method that makes use of Pascal's principle in measuring the pressure within the arterial walls based on the amount of external force the device exerts on the measurement area with the artery underneath. Data obtained from the pressure sensor is then sent to a computer for evaluation to determine the pressure readings.[80] Arterial tonometry does not require medical supervision for use and it also measures beat to beat pressure values, however it is dependent on the amount of force applied on the skin and requires continuous control of the control pressure and placement. The measurement is also highly dependent on the measurement site, being sensitive to skin abnormalities, and is also affected by motion artifacts. Accuracy of the observations and comfort of the patient also decreases with the time of monitoring. [78]

The volume clamp method is another continuous method that clamps the finger such that it would still provide a continuous flow of blood through the blood vessel in between. Blood volume is kept constant in the artery during each pulse as the external clamp would provide fluctuating pressure on the finger according to the volume, which then correlates to the changes in blood pressure within the artery.[81] This method also doesn't require the need for medical supervision and is a continuous measurement method for blood pressure while being partly occlusive. However as it is a clamping method, the vessel under the cuff is slightly compressed, which can affect blood circulation in the finger after long periods. It also requires continuous control of pressure in the clamp or cuff, which also results in reduced comfort. [78]

Pulse wave velocity is a method that involves two devices at a distance apart from each other along an arterial tree, where the time taken for one pulse to travel from one device to the other would be taken which would then be used to determine the blood pressure within an individual.[82] This method is a cuffless method, which is relatively more comfortable for users, and there's no need for medical supervision too. It also measures beat to beat pressure values and specific blood pressure values can be evaluated from standard biological signals. It can also be used during normal daily activities, although there's a need for precise measurement distance between the two sensors. The sensors also need occasional recalibrations and require initial supervision of medical professionals to determine the locations of arteries that can be used for measurement, affecting the accuracy of measurements. [78]

In the development of non-invasive wearable blood pressure sensors, a few criteria are very important. Firstly, the device developed has to be comfortable to users so that users are able to wear the device for a long period of time. One way to make this more comfortable for users is to adopt a cuffless method. The device should also not require supervision from the doctor or any medical personnel. Another criteria would be the accuracy of readings,

**Table 4. Accuracy categorization of wearable cuffless blood pressure sensor devices
based on IEEE Std 1708[83]**

Mean Absolute	Blood pressure	
Difference Grade	readings	ANSI/AAMI SP10
A	≤ 5 mmHg	pass
B	$\approx 5-6$ mmHg	pass or fail
C	$\approx 6-7$ mmHg	mostly fail, less pass
D	≥ 7 mmHg	fail

which would be in adherence to the global standards like the IEEE Standard for Wearable, Cuffless Blood Pressure Measuring Devices (IEEE Std 1708)[83].

Judging from the pros and cons of each method, the tonometry method and pulse wave velocity method are the only two methods that are suitable for use in cuffless approaches. However, the tonometry method requires a precisely controlled pressure exertion from the device which is very critical and the pulse wave velocity method is mainly disadvantaged by the distance measurement between the two measurement sites. Based on these findings, it is considered that the either of these two methods would be used in the design of the device. A more detailed description of both methods are given below in the next chapter.

2.2.4 OPTICAL FIBER IN BLOOD PRESSURE MEASUREMENT

Optical fibers have been widely used in the area of biomedical engineering. They are flexible, allowing easy insertions through the use of needles and catheters, enabling localized measurements within blood and tissues. The materials used for the manufacturing of these fibers are also non-toxic and biocompatible, making it safe for patients, and even when multiple fibers are close, there is low attenuation and no crosstalk between each fiber, making it a very good choice for application in surgeries or medical devices. Various types of sensors have been developed using this method, like the fabry-perot pressure sensors, blood flow sensors like

photoplethysmography devices, radiation dose sensors, etc. and in the area of bio-imaging, bioluminescence tomography, diffuse optical tomography and also optical coherence tomography. With respect to blood pressure sensors, the fabry-perot pressure sensor is very widely researched, however for it to be in the form of a wearable device, the optical fiber would have to be left perpendicular to the skin surface, which would cause major hindrances. On the other hand, the photoplethysmography device stands to have room for innovation given the complex reflective head at the tip of the sensor. Also, the optical coherence tomography method would also seem to be possible to innovate further for use as a blood pressure sensor. Thus, these two methods would be discussed further below.

Figure 20. Schematic diagram of an endoscope OCT device showing the 90 degrees scanning probe due to the position of the mirror [84]

Figure 21. LED and photodetector (PD) placements for transmission (left) and reflectance (right) PPG sensors[85]

Optical coherence tomography (OCT) is the real time tomographic imaging of biological tissues with high resolution achieved through the interference and back-scattered light received from tissues through the use of a broadband light source.[86, 87] This method is usually used for capturing internal organ tissues like during an endoscope. Through the use of MEMS techniques, endoscopic OCTs now have rapid scanning with even smaller sizes, increased reliability and flexibility in terms of scanning pattern. OCT makes use of both types of interferometry in its design, the frequency domain and time domain OCTs.[88] This concept would be a good method to be used for blood pressure sensing as it would firstly be able to detect the motion of blood within the blood vessels underneath the skin, and secondly, the device would be able to detect the blood pressure while the device is placed flat on the object surface (See Figure 20).

Photoplethysmography (PPG) is a method that is commonly found in the detection for blood pressure and it is a relatively mature technique in this field. Devices implementing this method are also commonly used as references for non-invasive blood pressure or heart measurement. The PPG signals reflect the blood movement inside the vessel beneath it and there are two different commonly used light wavelengths for such devices. The green light would exhibit larger intensity variation as compared to the infrared light, as the wavelength of light is most absorbed by red blood cells.[89, 90] There are two types of PPG sensors, the transmission type and the reflectance type (See Figure 21). The transmission type requires the measurement site to be relatively thin so that the transmitted light would be able to be captured by the photodetector placed on the other side. The reflectance mode makes use of the reflected light from the skin and blood vessels to determine the blood pressure waveform and it is affected by motion artifacts and other disturbances like slight movements of the measurement site and respiration. Other factors like the contact force of the probe, and measurement site itself also do play a part in the sensitivity of the device. The penetration depth

Figure 22. Schematic diagram of a holographic interferometry PPG sensor where object 32 is the reflective mirror, 34 is the optical fiber, 36 is a compound microscopic lens and 37 is a light diffuser.[91]

was previously determined by Huang et al [73] to be approximately 0.2mm for 463nm blue light, 0.4mm for 543nm green light, 0.5mm for 571nm yellowish green light and 0.62mm for 634nm red light. Through this concept, the idea of the 90 degrees exposure and the different penetration depths will be used.

A patent filed in 2011 by Bladwin et al [91] has made use of the 90 degrees exposure concept with the optical fiber exposing the mirror with light which is reflected to the skin and back into the optic fiber through the same path (See Figure 22). This concept applies the use of holographic interferometry which is also used for other kinds of non-destructive vibrational analysis. By building up on this concept, and including a new innovation, the new device would be cheaper with less components in each device. This new innovation would be in the aspects of modification of the optic fiber tip, which would effectively remove the need for the reflective mirror and compound microscopic lens.

2.2.5 BALL-SHAPED SIDE-FIRE OPTICAL FIBER

Studies have been conducted to determine the beam profiles of light exiting a new ball-shaped optic fiber tip (See Figure 23). This new type of fiber tip allows self-focusing of the beam without the need for additional optical lens. On top of this, by slicing the circular tip at

an angle and polishing the cut surface, the cut surface acts as a reflective surface that diverts the light path at an angle to the original light path (See Figure 24). This is the ball shaped side-fire optical fiber tip as mentioned by Cappius et al [92]. The research presented by them confirmed the method works and on top of that, the light reflected from an object surface is able to enter back into the optical fiber through the same fiber tip. The paper however did not mention any optimization of the slicing angle or any dimensional measurement of the focal distance. The optical fiber material used is quartz, where the total internal reflection angle is from 43 degrees onwards.

From the literature findings, it is conclusive to say that this type of optical fiber tip would be able to work effectively if incorporated into a wearable blood pressure device, eliminating the need to multiple parts in a device. This would greatly reduce costs and also assembly and naturally the added risks with it. With these set of information, the theories behind the pulse wave velocity and arterial tonometry will be further mentioned in the next chapter, which would be the main theories used in the development of device design concept.

Figure 23. Beam profiles for different types of optical fiber tips[93]

Figure 24. Illumination of surface (left) showing the ball-shaped side-fire optic fiber and the shape of the optic fiber tip(right)[92]

CHAPTER 3: THEORY

3.1 SEISMIC MOEMS DEVICE

3.1.1 SPRING-MASS-DASHPOT SYSTEM

Figure 25. Schematic diagram of accelerometer[94]

The basic principle of the device works based on the spring-mass-dashpot system, where the springs are the beams connecting the proof mass to the frame and damper is the thin film of air between the frame and mass (See Figure 25). When force is applied unto a proof mass supported by springs, according to Newton's Second Law, the acceleration of a body would be directly proportional and in the same direction as the net force acting on a body while being inversely proportional to its mass.[94] This leads to a series of calculations that would then lead to the current inertial motion equations used for most accelerometers, where

$$\sum F_{acting\ on\ mass} = 0 \quad (1)$$

Calculations of natural frequency and resultant displacement of mass would be calculated according to the theoretical equations given below.[94]

EOM:
$$m\ddot{x} + c\dot{x} + kx = F_o \sin \omega t \quad (2)$$

For steady state response, let $x = X \sin(\omega t - \phi)$.

Thus,
$$\ddot{x} = -\omega^2 X \sin(\omega t - \phi + \pi), \quad (3)$$

and
$$\dot{x} = \omega X \cos(\omega t - \phi) = \omega X \sin(\omega t - \phi + \frac{\pi}{2}) \quad (4)$$

Inertia force,
$$m\ddot{x} = m\omega^2 X \sin(\omega t - \phi + \pi) \quad (5)$$

Damping force,
$$c\dot{x} = c\omega X \sin\left(\omega t - \phi + \frac{\pi}{2}\right) \quad (6)$$

Spring force,
$$kx = kX \sin(\omega t - \phi) \quad (7)$$

$$(k - m\omega^2)X \sin(\omega t - \phi) + c\omega X \cos(\omega t - \phi) = F_o \sin \omega t \quad (8)$$

By re-expressing the RHS of equation as sine function and equating amplitude of sine function on both sides of the equation,

$$X\sqrt{(k - m\omega^2)^2 + (c\omega)^2} = F_o \quad (9)$$

$$X = \frac{F_o}{\sqrt{(k - m\omega^2)^2 + (c\omega)^2}} \quad (10)$$

And
$$\phi = \tan^{-1}\left(\frac{c\omega}{k - m\omega^2}\right) \quad (11)$$

$$\omega_n = \sqrt{\frac{k}{m}} \quad (12)$$

For the mathematical equations, m is the mass of the proof mass, c is the damping constant, k is the spring constant, X is the amplitude of motion of the mass, ω is the induced frequency of the system, and ω_n is the natural frequency of the system.

3.1.2 DESIGN CONCEPT

The proposed device comprises of a silicon based SOI MEMS layer attached onto a CMOS layer where the electrical outlay and phototransistors would be fabricated on. Design concept would be based on the optical grating method of detection, whereby changes in forces would be detected through the differential effect on the light intensity captured by the photodetector due to the direct flooding of proof mass using incident light as mentioned in the previous chapter (See Figure 26). The design of the MEMS layer proof mass would be developed and fabricated in-house. The CMOS layer would consist of strips of phototransistor areas that would detect the displacement of the silicon proof mass through the increase/decrease of light intensity upon each phototransistor. Figure 27 shows a demonstration of how the device would work using the abovementioned method of detection with the inclusion of gratings on

Figure 26. Basic concept of device (a) Overall design, (b) Brief design of MEMS layer, (c) Initial design of CMOS layer without electrical outlay[95]

Figure 27. Method of detection for designed device[95]

the proof mass. Gratings are etched into the proof mass via the handle layer to allow light to pass through from the LED above the proof mass to the phototransistor portion under the proof mass.

Upon the acceleration of the device, inertial forces experienced by the proof mass would cause the proof mass to be displaced in the opposite direction of the acceleration. This deflection would then adjust the amount of light incident on the phototransistor placed below the proof mass. This measure of light intensity would then be converted into an electrical signal that would be sent into a processor where it would then convert the signal into an output reading. Through this, different measures of light intensity would be captured and compared with reference to one another to finally determine the actual displacement read by the device.

3.1.3 DESIGN VARIABLES

An overview of the basic design is given in Figure 28. This device requires the use of both top-side etching and bottom-side etching, where a huge source of error would be due to the angular misalignment of both masks with the wafer during fabrication procedures. A study of the effects of misalignment between two layers was done and it is observed that a maximal tolerance for error would be $\pm 2.5^\circ$ for a 1% chance of error and $\pm 8^\circ$ for a 10% chance of error (See APPENDIX C). On top of these calculations, with current technological standards in DRIE and photolithography procedures, it is safe to say that the chances of misalignment are very low.

Among the variables mentioned in Figure 28, the thickness values for the device will be according to the thickness values given by the SOI wafer which would be fixed. The overall device area dimensions would also be fixed according to the dimensions of the LED and phototransistor packages used. For the gaps and gratings in the device, the widths are standardized throughout the whole device to facilitate easier fabrication. So far, all the variables mentioned above are kept at constant values throughout the development of designs for easier

comparison during the evaluation of design prototypes. With regards to the other variables that are adjusted during optimization simulations, it is observed that the major dimensions that affect the displacement and natural frequency of the device would be the spring constant, k , and the actual mass of the proof mass, m_{mass} . These two factors are affected by the variables given in Figure 28 as follows:

$$k = 4k_{beam} \quad (13)$$

$$k_{beam} = \frac{12EI}{B_L^3} \quad (14)$$

$$I = \frac{bh^3}{12} = \frac{t_{dev}B_W^3}{12} \quad (15)$$

$$\therefore k = \frac{4Et_{dev}B_W^3}{B_L^3} \quad (16)$$

Assuming mass of Silicon Oxide layer is negligible as thickness of BOX layer is very small,

$$m_{mass} = \rho \times V_{mass}$$

$$m_{mass} = \rho[(M_L \times M_W - G_W \times G_L \times N)t_{sub} + (M_L \times M_W - (G_W \times N + S_W \times (N - 1)) \times G_L)t_{dev}] \quad (17)$$

$$X = \frac{F_o}{\sqrt{(k - m\omega^2)^2 + (c\omega)^2}} = \frac{F_o}{\sqrt{\left(\frac{4Et_{dev}B_W^3}{B_L^3} - m_{mass}\omega^2\right)^2 + (c\omega)^2}} \quad (18)$$

$$\omega_n = \sqrt{\frac{k}{m_{mass}}}$$

$$= \sqrt{\frac{\frac{4Et_{dev}B_W^3}{B_L^3}}{\rho[(M_L \times M_W - G_W \times G_L \times N)t_{sub} + (M_L \times M_W - (G_W \times N + S_W \times (N - 1)) \times G_L)t_{dev]}}} = \sqrt{\frac{4Et_{dev}B_W^3}{\rho[(M_L \times M_W - G_W \times G_L)t_{sub} + (M_L \times M_W - (G_W + S_W) \times G_L)t_{dev}]B_L^3}} \quad (19)$$

By equating the variables together, a relationship can be found for the optimization of the design. The three main variables are the **Mass of the Silicon** for the proof mass, **Beam Width** and **Beam Length**. Simulations are then carried out to evaluate the effects of each variable to produce a list of designs in accordance with the requirements given in Chapter 1.2.

Figure 28. Variables of interest for device (a) Top view, (b) Sectional view, (c) Isometric view of square hole in device layer with table of variables

3.2 WEARABLE OPTICAL BLOOD PRESSURE SENSOR

3.2.1 PULSE WAVE VELOCITY

Figure 29. Simulated model of artery used to determine Pulse Wave Velocity. [96]

Pulse wave Velocity, or also commonly called pulse transit timing (PTT) method can be explained using arterial wave propagation to relate wall elasticity to PTT.[97] The concept works based on the detection of a single pulse at two different locations across a vessel, as given in Figure 29. The time lag between the detection of that same pulse is thus the PTT. This method neglects the viscous effects of blood and arterial walls which have relatively small effect in central and peripheral arteries. Apart from these variables, another factor that quantifies the pressure and volume relationship is compliance, C , which can be given in terms of area compliance C derived from a combination of Laplace's law for cylindrical tube,[98]

$$C = \frac{\Delta A}{\Delta P} \quad (1)$$

where

$$C = \frac{3\pi r(r + h)^2}{E(2r + h)h} \quad (2)$$

which can subsequently be simplified into

$$C = \frac{2\pi r^3}{E \cdot h} \quad (3)$$

and a relationship between the wall thickness h , the internal radii r , and the elastic modulus of the wall, E , which is assumed to be non-constant within a region of concern, A , as such.

$$E(P) = E_o e^{\alpha P} \quad (4)$$

A relationship between pressure and arterial cross sectional area has been developed by Langewouters et al [99], given as follows

$$A(P) = A_m \left(\frac{1}{2} + \frac{1}{\pi} \tan^{-1} \left(\frac{P - P_o}{P_1} \right) \right) \quad (5)$$

$$dA = \frac{A_1 \cdot dp}{1 + \left(\frac{P - P_o}{P_1} \right)^2} \quad (6)$$

$$A = A_1 \cdot \tan^{-1} \left(\frac{P - P_o}{P_1} \right) + A_2 \quad (7)$$

where A_m is the maximal cross-sectional area of the aorta at high pressures, P_o is the transmural pressure at the inflection point at of the pressure-area curve, P_1 is the steepness of the rise of the curve. These three variables are user dependent variables.

Also, for the boundary condition $A(p = -\infty) = 0$,

$$A_2 = \pi \frac{A_1}{2} \quad (8)$$

Thus, the static compliance of aorta is given as

$$C(P) = \frac{A_m}{\pi P_1 \left[1 + \left(\frac{P - P_o}{P_1} \right)^2 \right]} \quad (9)$$

Assuming that there is negligible variations in region A of by applying conservation of mass and momentum, a set of simultaneous equations are derived as follows,

$$Q(x) - Q(x + dx) + \frac{d(Adx)}{dt} = 0 \quad (10)$$

$$\frac{\partial Q}{\partial x} + C(P) \frac{\partial P}{\partial t} = 0 \quad (11)$$

$$A[P(x) - P(x + dx)] = \rho dx \frac{dQ}{dt} \quad (12)$$

$$\frac{\partial P}{\partial x} + L \frac{\partial Q}{\partial t} = 0 \quad (13)$$

where x and t are referring to positioning and time, Q is the volume flow rate, ρ is the density of blood and $L = \rho/A$ which is the arterial inertance per unit length. Using these two equations, the pressure function $P(x, t)$ can be written as below,[97]

$$P(x, t) = f\left(x \pm \frac{t}{\sqrt{LC(P)}}\right) \quad (14)$$

The time taken for pulse travel is thus given as,

$$PTT = l\sqrt{LC(P)} \quad (15)$$

and subsequently the pulse wave velocity is,

$$PWV = \frac{1}{\sqrt{LC(P)}} \quad (16)$$

where PTT is inversely related to BP, and also directly affected by smooth muscle contractions, distance from heart, heart rate and age. The smooth muscle contractions have been reported to have an effect on the static pressure-area relationship, causing an increase in the pressure levels by around 10 to 30 mmHg for a certain area. The distance from the heart would affect the blood pressure waveform as given in Figure 15, thus if the distance was less, the waveform would then be less amplified. The PWV equation is also commonly provided as the Moens-Korteweg equation form,

$$PWV = \sqrt{\frac{E \cdot h}{2r\rho}} = \sqrt{\frac{E_0 e^{\alpha P} \cdot h}{2r\rho}} \quad (17)$$

with E_0 as the zero pressure modulus, α a parameter depending on the vessel itself, e being Euler's number of 2.718, h is the wall thickness, r the radius of the vessel and P the blood pressure. By using these set of information, PTT can be used to monitor the blood pressure.

3.2.2 ARTERIAL TONOMETRY

The method of tonometry as previously mentioned employs Pascal's principle to measure the pressure in the artery based on the external force exerted on the artery wall over the compressed area.[80] Pascal's principle states that the pressure applied to an enclosed fluid is transmitted throughout every part of the fluid and its vessel walls without diminishing. In specifics, the circumferential tension of the inner vascular wall changes while the external pressure flattens the internal arterial wall. Upon a certain level of external pressure, the intra-arterial pressure is then equal to the external pressure which makes it possible to be measured by external pressure.[28] It was also found that arterial waveforms measured at various physiological locations, eg. brachial artery, radial artery, etc., were able to be used for analyzing important

Figure 30. Schematic diagram of an arterial tonometer (left) and the detected observations (right)[100]

hemodynamic parameters.

There are two typical types of tonometers, the first being one with a flexible diaphragm, and another that has a pressure transducer that is directly in contact with the skin. The flexible diaphragm method has a device that consists of a fluid that is covered by a diaphragm layer that would be in direct contact with the skin. By maintaining the equilibrium within the

enclosed fluid compartment through minor adjustments in the volume, the changing arterial pressure can be detected. The second method uses pressure transducers that are either capacitive or piezoresistive for detection and the entire sensor itself has to be flat and relatively stiff and small. Both methods been commonly used for evaluating hypertension, arterial stiffness, augmentation index, heart failure and response of the central blood pressure system in response to different cardiovascular stimulants or drugs. A schematic diagram of a tonometer is given in Figure 30.

According to Holzapfel et al [59], several models have been developed to provide estimations of the deformations in an artery, with different assumptions and approximations to account for different types of analyses. For simplification of understanding the effects of blood

Figure 31. Schematic diagram of a blood vessel

pressure on wall deformation, a vague correlation of both variables is given according to the Young-Laplace equation, with the assumption that the thickness of the wall is small as compared to the diameter of the lumen (See Figure 31). The cylinder stress is given as

$$\sigma_{\theta} = \frac{Pr}{t} = \frac{F}{tl} \quad (18)$$

$$F = Plr \quad (19)$$

$$-kX = Plr \quad (20)$$

$$X = \frac{-Plr}{k} \quad (21)$$

where P is the blood pressure, r the inside radius of the blood vessel, t the thickness of the vessel wall, F the circumferential force, l the axial length of the vessel, and X the displacement of the vessel wall. In a recent study by Loh et al [101], it was observed that the mean right arterial diameter for the population of Singapore was $2.45 \pm 0.54\text{mm}$. Through this theory, the displacement that is captured by the tonometer would then be able to detect the blood pressure within the vessel. One advantage of using the tonometer is that it only requires a single device to detect the blood pressure, however this method requires regular calibration of the device.

CHAPTER 4 : SIMULATION RESULTS

4.1 SEISMIC MOEMS DEVICE

To ensure that the results obtained from simulations are accurate, two different approaches are carried out, namely the Finite Element approach and Numerical approach. Results obtained from both simulations are then compared against each other for accuracy before being used for further optimization of designs.[102] In the initial stages, Finite Element Simulations are first used to determine the effects of each individual variable and their interrelations with one another with respect with the output of the device. By studying the trends observed, it is then possible to develop a final device with optimized output that's within expectations. As a reiteration, the output of the device is targeted to display a minimum of 0.1 microns displacement upon an external force of 2G (where $G=9.81\text{m/s}^2$) with the 1st mode natural frequency at approximately 2000Hz.

Throughout the simulations carried out, the constant variables, $D_L \times D_W$, are according to the LED and phototransistor dimensions obtained from the datasheets provided (See APPENDIX D and APPENDIX E), and thickness values are according to SOI wafer dimensions given. Finite Element Simulations were carried out using the CAD software, ANSYS Workbench 14.0, available in the CAE Lab 1. Numerical Simulations were obtained by applying the theories mentioned in CHAPTER 3. The constants used for the simulations are given in Table 5.

Before the simulations were carried out, a short simulation on the different frequency modes that would occur during acceleration was carried out to determine the frequency mode that is of interest. This is to check to determine if there would be any contamination of results with other unwanted frequency modes, as logically there would be multiple modes of vibration in a device at different points of time. By checking the frequencies of each mode of vibration, it would be possible to check on the possibility of cross axis contamination of results. The

Table 5 . Device area and SOI wafer specifications

SOI wafer specifications:		
Device layer thickness	t_{dev}	30 μ m
Oxide layer thickness	t_{SiO_2}	1 μ m
Handle layer thickness	t_{sub}	500 μ m
Diameter of wafer		4 inch
Device area specifications:		
Device length	D_L	2350 μ m
Device Width	D_W	3300 μ m

Figure 32. Number of frequency modes simulated and their corresponding frequencies.

further the frequencies are from each other, it means that there would be less chances of having the proof mass moving in both modes of vibration at the same time. The test run simulated six frequency modes as shown in Figure 32, and the results show that the in-plane mode of vibration, which is the mode of interest, is the 1st mode of vibration. It is also observed that the frequency of this first mode is relatively far away from all the other modes of vibration,

implying that there would be little effect of the other modes of vibration when the designed device is being used. This would be vital to the project and therefore, all frequencies measured in this project would be in relation to the 1st mode of vibration of the proof mass.

4.1.1 PARAMETERS STUDY USING FINITE ELEMENT SIMULATION

Figure 33. Effects of Beam Length variation on frequency (left) and displacement (right)

Figure 34. Effects of Beam Width variation on frequency (left) and displacement (right)

Figure 35. Effects of Proof Mass Area variation on frequency (left) and displacement (right)

Preliminary studies of individual variables and their effects are seen in the following graphs. The three main variables would be the beam length, beam width and area of proof mass. Figure 33 to Figure 35 above show that displacement is increased under the following conditions:

- a) Beam length increases;
- b) Beam width decreases;
- c) Proof mass area increases.

On the other hand, natural frequency of the device is increased when

- a) Beam length decreases;
- b) Beam width decreases;
- c) Proof mass area decreases.

Thus, it can be seen that there is a delicate balance between the natural frequency and maximum displacement experienced under a certain force and by optimizing these three variables, a balance can be reached that is within the requirements set in this project. To determine the effects of the interaction of all three variables, graphs are plotted out using a

Figure 36. Graphs for Beam Length, Beam Width and Area variation with respect to frequency and displacement

Figure 37. Zoom in graphs for Beam Length (top left), Beam Width (top right) and Mass Area (bottom) variation

combined factor, $Pro \quad Mass \ Area \times \left(\frac{Beam \ Length}{Beam \ Width}\right)^3$, as according to the theoretical impact of each variable on the resultant output. At every point in the simulation, two variables are kept constant while one is varying and the effects of each variable can be seen in Figure 36. Through this, the overall effects of all three variables can then be studied. The additional zoom in portions of the figures show some of the points simulated in the FEM simulation that meet the criteria given and can be used as one of the device designs in the next chapter (See Figure 37). As mentioned previously the resonant frequency and displacement are inversely proportional to one another,

4.1.2 COMPARISON AGAINST NUMERICAL SIMULATION RESULTS

The results of both the Numerical Simulation and Finite Element Simulation are given in Table 6, where the series of devices were obtained after optimization of the design variables. It is found that there is a maximum of 5% deviation of FEM results for the 1st mode natural

Table 6. Summary of Numerical Simulation and FEM Simulation Results

	Natural Frequency (Hz)			In-plane displacement (μm)		
	Theory	FEM	% Diff	Theory	FEM	% Diff
Device A	2967.231	3094.221	4.280	0.056	0.052	-7.236
Device B	2001.872	2100.675	4.936	0.124	0.114	-8.241
Device C	2035.329	2084.215	2.402	0.120	0.116	-3.625
Device D	2354.687	2397.502	1.818	0.090	0.088	-2.279
Device E	2013.172	2094.619	4.046	0.123	0.114	-6.692
Device F	2253.694	2265.076	0.505	0.098	0.098	0.148
Device G	2076.411	2106.509	1.450	0.115	0.114	-1.364
Device H	1930.600	1971.612	2.124	0.133	0.130	-2.796

Table 7. FEM Simulation Results of First Three Modes of Resonance

	Mode 1 (Hz)	Mode 2 (Hz)	Mode 3 (Hz)
Device A	3094.22	14005.57	29711.62
Device B	2100.67	9593.11	24681.22
Device C	2084.21	8436.37	23597.69
Device D	2397.50	8483.49	27781.08
Device E	2094.62	9385.09	25082.21
Device F	2265.08	8944.57	26855.40
Device G	2106.51	8342.36	24247.61
Device H	1971.61	7773.36	29013.42

Figure 38. Comparison of project device with other similar MOEMS motion detection devices previously mentioned

frequency across all devices. For the in-plane displacement, there is a maximum of 8.24% difference when the difference for the natural frequency is also high. These differences could be due to the limitations of the ANSYS Educational License, which only allowed coarse meshing of the device resulting in less accurate data. Although the errors, it is observed that both numerical simulation and FEM simulation results are within the range that have been set above and can be used as benchmarks during the testing of the device prototype. Table 7 displays the FEM simulations of the first three modes of resonance showing that there would be no effects from the other modes of vibration within the range of 0 to 500 Hz. By using an average value of the simulated in-plane displacement, the sensitivity is calculated to be 92.7 nm/G with brownian noise equivalent acceleration to be approximately $25.3nV/\sqrt{Hz}$. This sensitivity places the simulated device to be positioned closer to the top right region as observed

in Figure 38, which depicts higher sensitivity with higher natural frequency, which is the aim of the project.

4.1.3 DEVICE DESIGN SPECIFICATIONS

Using results given in Table 6, the eight devices with variations in each parameter were designed for fabrication (See Table 8). The main purpose of using eight different designs would be to test the fabrication procedures and also the survivability of the devices during the handling stages of integration and assembly. The outputs of each device are all relatively similar in terms of natural frequency and displacement under 2G force, thus an added criteria to the selection of optimized device would be the ease of fabrication. The differences between each device are highlighted in red in Table 8.

Using these specifications, 5" glass masks were designed and sent for fabrication. As the fabrication of the device calls for both top side etching (from device layer) and bottom side etching (from handle layer), two masks were needed for the fabrication process.

Table 8. Summary of specifications for each design

Design		Specifications
A		Device Area ($D_W \times D_L$): 3.3mm x 2.35mm
		Mass Area ($M_W \times M_L$): 2mm x 1mm
		Beam Length (B_L): 200 μ m
		Beam Width (B_W): 6 μ m
		Gap Width (G_W): 30 μ m
		Grating Length (G_L): 900 μ m
		Dist btwn gratings: 45 μ m
		Number of gratings: 25

B		<p>Device Area ($D_W \times D_L$): 3.3mm x 2.35mm</p> <p>Mass Area ($M_W \times M_L$): 2mm x 1mm</p> <p>Beam Length (B_L): 260μm</p> <p>Beam Width (B_W): 6μm</p> <p>Gap Width (G_W): 30μm</p> <p>Grating Length (G_L): 900μm</p> <p>Dist btwn gratings: 45μm</p> <p>Number of gratings: 25</p>
C		<p>Device Area ($D_W \times D_L$): 3.3mm x 2.35mm</p> <p>Mass Area ($M_W \times M_L$): 2mm x 1mm</p> <p>Beam Length (B_L): 300μm</p> <p>Beam Width (B_W): 7μm</p> <p>Gap Width (G_W): 30μm</p> <p>Grating Length (G_L): 900μm</p> <p>Dist btwn gratings: 45μm</p> <p>Number of gratings: 25</p>
D		<p>Device Area ($D_W \times D_L$): 3.3mm x 2.35mm</p> <p>Mass Area ($M_W \times M_L$): 2mm x 1mm</p> <p>Beam Length (B_L): 300μm</p> <p>Beam Width (B_W): 8μm</p> <p>Gap Width (G_W): 30μm</p> <p>Grating Length (G_L): 900μm</p>

		Dist btwn gratings: 45μm Number of gratings: 20
E		Device Area ($D_W \times D_L$): 3.3mm x 2.35mm Mass Area ($M_W \times M_L$): 2mm x 1.8mm Beam Length (B_L): 210μm Beam Width (B_W): 6μm Gap Width (G_W): 30μm Grating Length (G_L): 1500μm Dist btwn gratings: 45μm Number of gratings: 25
F		Device Area ($D_W \times D_L$): 3.3mm x 2.35mm Mass Area ($M_W \times M_L$): 2mm x 1.8mm Beam Length (B_L): 220μm Beam Width (B_W): 7μm Gap Width (G_W): 30μm Grating Length (G_L): 1200μm Dist btwn gratings: 45μm Number of gratings: 25

G		<p>Device Area ($D_W \times D_L$): 3.3mm x 2.35mm</p> <p>Mass Area ($M_W \times M_L$): 2mm x 1.8mm</p> <p>Beam Length (B_L): 240μm</p> <p>Beam Width (B_W): 7μm</p> <p>Gap Width (G_W): 30μm</p> <p>Grating Length (G_L): 1500μm</p> <p>Dist btwn gratings: 45μm</p> <p>Number of gratings: 25</p>
H		<p>Device Area ($D_W \times D_L$): 3.3mm x 2.35mm</p> <p>Mass Area ($M_W \times M_L$): 2mm x 1.8mm</p> <p>Beam Length (B_L): 220μm</p> <p>Beam Width (B_W): 7μm</p> <p>Gap Width (G_W): 30μm</p> <p>Grating Length (G_L): 1200μm</p> <p>Dist btwn gratings: 45μm</p> <p>Number of gratings: 20</p>

4.1.4 MASK DESIGN SPECIFICATIONS

Figure 39. Overall design layout of wafer masks

Overall design of the mask layout is given in Figure 39 seen below. The masks are designed using dark field mode and the gap between each device on each mask is $400\mu\text{m}$ to allow enough space for the dicing saw to pass through during wafer dicing. The dicing saw alignment marks are the cross shaped marks seen above in Figure 39 in between each proof mass. The dicing saw would also cut out the shape of the frame of the proof mass. The 5mm gap on the circumference of the wafer is given to provide ample space for the protrusion pins of the DRIE machine which is required in one of the processes for DRIE. Also, wafer alignment marks are placed 35mm away from the center of the wafer. These alignment marks are used for the alignment of the wafer to the mask before each photolithography process. For each device, there are a total of 75 units fabricated using a single SOI wafer, to ensure that a satisfactory number of devices are produced during each fabrication. Table 9 below shows the

mask designs for each device. Specifications used for each design are according to the specifications mentioned previously after optimization of initial device design.

Table 9. Mask designs for individual devices

	Top Mask	Bottom Mask
Design A
Design B
Design C

Design D
Design E
Design F

4.2 WEARABLE OPTICAL BLOOD PRESSURE SENSOR

4.2.1 PROPOSED APPROACHES FOR DEVICE

A few device approaches were developed for the wearable optical blood pressure sensor. These devices make use of the theories mentioned in the previous chapter. These different approaches employ the use of optical methods to obtain the amount of deflection on the skin due to pressure pulses from the arteries. The artery of interest is the left radial artery that is on the left wrist, in the position where the watch buckle is usually at. The size of the sensing element of the device would be limited to within 2cm x 1cm x 1.5cm maximum. The designs

were developed with the help of optical simulations that would determine different dimensions of the design.

APPROACH A

Figure 40. Design approach of new optical blood pressure device

Figure 41. Placement of devices on wrist

For the blood pressure sensor, a proposed approach for an optical device is developed using a tonometric method (See Figure 40). The approach makes use of a mechanical setup whereby the detection is determined by the intensity of light that enters the phototransistor from the VCSEL light source that are set on a single printed circuit board. The light is transferred through an optical waveguide that is made using PMMA. The PMMA is cut to form a U shape, where one end is fixed on the VCSEL light source, and the other is a flexible arm which is directly above the phototransistor. The flexible arm is not fixed over the phototransistor, and is placed over the surface of skin that is used for detection. Light is passed

from the VCSEL through the optical waveguide unto the phototransistor. It was previously mentioned that blood pressure within the radial artery would cause vibrations that can be detected on the skin surface. By placing the device on the skin surface, the device would then be able to detect the vibrations through the displacement of the optical waveguide flexible arm. As the arm of the waveguide displaces, the light intensity at the phototransistor side would change consequently, and thus allows the detection of the blood pressure. Two devices are used for this purpose as one device would only be able to obtain the pulse rate. Two devices would enable the measurement of the blood pressure values through the application of the pulse wave velocity method. The two devices would be placed along the radial artery at a small distance apart as seen in Figure 41. The flexible substrate in the figures can be a plaster which can be used to stick unto the skin surface.

APPROACH B

Figure 42. Schematic diagram of Device Approach B (left) and placement on wrist (right)

This approach employs the use of two devices at two different measurement sites that would be used in the calculation of blood pressure values using the pulse wave velocity method (See Figure 42). It would resemble the piezoresistive tonometry method, except that the detection mechanism would be using optical means. This method also uses the PWV method to eliminate the need for constant pressure exertion for the device to work effectively. An IR-LED will be used to illuminate both devices through the use of the ball-shaped side-fire optical fiber tip. A thin reflective membrane will be used as an interface with the skin surface to prevent contamination of the device. When the device is placed over one measurement site, the blood pressure waveform is captured through the motion of the blood vessel walls which can be observed from the skin layer. The two waveforms from the two devices can then be put together to determine the pulse wave velocity and subsequently the blood pressure values. The readings captured from the measurement sites would be susceptible to motion artefacts, however that would not be of concern as the device would only require the PTT to obtain the pressure values.

APPROACH C

Figure 43. Schematic diagram of Approach C (left) and placement on wrist (right)

Approach C relies on the usage of two different wavelengths of light, the 750nm Red light and 1550nm IR light for measurements. The approach is similar to the photoplethysmography method (See Figure 43). The Red light will penetrate the upper layers of the skin to the blood vessel walls and be reflected back into the optic fiber from it (See Figure 17). The IR light would be reflected from the upper skin layer back into the optic fiber above. The entire singular device works like a Michelson interferometer where the reference mirror is fixed and the light captured back into the optic fiber is computed into displacement measurements via fourier transform. The frequency shift for both the IR light and Red light would then determine the displacement measurements for the skin layer and the blood vessel walls. By taking into consideration both displacements, the motion artefacts due to reasons like respiration or hand motion can then be eliminated as they would all be caught in the readings for the IR light. For the blood pressure values though, calibration would need to be carried out before the actual usage of the device, which means that another blood pressure device would be required for initial use. Through this calibration, the device would then be able to determine the blood pressure values accurately, and for this approach, only one device would be needed.

Three design approaches have been developed for the wearable blood pressure sensor, each with its pros and cons. Approach A makes use of a simple fabrication of a PMMA optical waveguide for each device. Although this method is cheap and easily fabricated using a laser cutter, different issues will arise like the surface finishing of the entry and exit points of the PMMA waveguide, resulting in the need for additional fine polishing procedures. As such, high error rates would be encountered. Approach B requires the use of two devices which cannot be too far apart in the arterial tree, however it is able to effectively show the blood pressure values directly without the use of any prior calibration. Approach C on the other hand requires the use of a precalibration using another blood pressure sensor, however it requires only one device which means that it would be cheaper and more user friendly. Accuracy of this design would require calibration. Both Approach B and C involves the novel application of the ball-shaped fiber-optic tip in wearable biomedical devices. With this, the next step would be to run optical and mechanical simulations to determine the effectiveness of these three design concepts.

4.2.2 SOLIDWORKS AND ZEMAX OPTICAL SIMULATIONS

The aim of the works mentioned in this section is to show the effects of different variables of the different approaches.

APPROACH A

Figure 44. Observations from a combination of Zemax and Solidworks simulations

Using this approach, the major consideration of the device would be the design of the optical waveguide. To optimize the design to produce the maximum effects during the displacement of the optical waveguide arm, SolidWorks and Zemax simulations were carried out to determine the best dimensions for the waveguide (See Figure 44). A systematic approach was used to obtain the values of important variables like the thickness of the waveguide, radius of the curve, and placement of the vessel along the waveguide arm. These variables are simulated based on exerting a pressure that is according to the maximum systolic blood pressure of 120 mmHg on an area of 10mm x 10mm of the optical waveguide.[103] The location of the 10mm x 10mm area is also varied along the length of the optic waveguide in accordance with the “placement” variable given in Table 10. Through the comparison of the different variables in Table 10, the set of variables with the highest difference in power is the one that would be considered for fabrication.

With these set of variables, the optical waveguide is cut out from a piece of PMMA using laser cutting. The waveguide is then polished before use for the next set of experiments.

At the same time, a rectangular waveguide using the same thickness, length and placement of blood vessel as the curved waveguide is also created for use as a comparison against the curved waveguide.

Table 10. Variables used for simulations

Length (mm)	Placement (mm)	Thickness (mm)	Outer Radius (mm)	Width (mm)	Simulated Deflection (mm)	Detector size (mm ²)	Before Power (W)	After Power (W)	Difference (mW)
20	1	3	5	10	0.0248413	10x3	0.54387	0.54263	1.24
20	15	3	5	10	0.00531122	10x3	0.54387	0.54375	0.12
20	1	3	4	10	0.0189703	10x3	0.435	0.4344	0.6
20	1	3	6	10	0.0189703	10x3	0.65218	0.65477	2.59
20	15	2	5	10	0.0211859	10x3	0.82617	0.82617	0
20	1	2	5	10	0.0915589	10x3	0.82617	0.82617	0
20	1	2	4	10	0.0713646	10x3	0.66093	0.66093	0
20	1	2	6	10	0.0189703	10x3	0.99142	0.99142	0
20	1	2	6	10	0.0189703	10x2	0.99142	0.93247	58.95
20	15	2	5	10	0.0211859	10x2	0.80877	0.80705	1.72
20	1	2	5	10	0.0915589	10x2	0.80877	0.7877	21.07
20	1	2	4	10	0.0713646	10x2	0.64693	0.63533	11.6

APPROACH B and C

Figure 45. Effects of ball radius on power detected (bottom) based on schematics given (top)

For Approach B and C, the main important portion for simulation would be regarding the design of the optical fiber tip itself. There are several variables that would need to be optimized for this fiber to work like the radius of the ball shape at the tip, the cutting angle and depth of cut for the ball shape itself too. The standard diameter for the optical fiber was kept at 2mm as there was no need for small flexible diameters that would create more errors. For the simulations, infrared was used as a light source at 850nm as this wavelength was able to penetrate through the dermal layer to the arterial walls and get reflected from there back into the optical fiber. As it is unable to simulate the transmission and reflectance properties of the skin layer, a reflector was used instead to standardize all the experiments.

In Figure 45, the effects of the ball radius were studied and the detected power from the fiber with ball radius of 1.7mm was observed to have the greatest change within the four different radii used in the simulations. This change would result in greater fluctuations during the detection. The ball radius was also kept to a minimum as the large diameters may cause strain and stress on the fiber itself and also weigh the tip down, causing an additional source of error.

Figure 46. Effects of change in cutting angle on detector output (bottom) based on schematics given (top)

In the previous chapter, it was mentioned by Laurent et al [70] that the estimated maximum deflections of the radial artery walls are 0.02mm tangentially. A series of angles were tested from 30 degrees to 60 degrees for this simulation. Any angle beyond 60 degrees would not be reflected down into the simulated skin surface and back into the fiber, thus no

further angles were considered. In Figure 46, the results have shown that the angle with the greatest change within any range would be the 60 degrees cut angle from 0.65mm to 0.8mm. Also, the 50 degrees cut shows a linear increase in the optical power detected, which also makes it very suitable for use. Depending on further criteria, either the 50 degrees or the 60 degrees angled cut would be the best for the device.

Figure 47. Effects of change in cutting depth on detector output (left) based on schematics given (right)

Simulations for the cutting depth of the ball shaped tip was carried out and the results have shown that with the cutting depth of 0.8mm, the change in the detector output was the highest when the reflector was in the 0.25mm to 0.55mm range (See Figure 47). This cutting depth is then used in the design of the optical fiber itself. Similarly, after all the simulations were carried out, the design variables for the optical fiber are given as follows.

Optical Fiber Diameter:	2 mm
Ball Shaped Tip Radius:	1.7 mm
Cutting Angle:	50 degrees
Cutting Depth:	0.8 mm

Figure 48. Comparison of ball radii dimensions on red and infra-red wavelength light.

A further confirmation was carried out for the ball shaped radius after obtaining both the cutting angle and depth, comparing between 1.5mm, 1.6mm and 1.7mm radii. Observations found that the slope for the 1.7mm radius ball shaped tip has the largest slope within the similar range of 0.25 to 0.55mm (See Figure 48). As previously mentioned in Chapter 2 that the approximate tangential deflection of the skin due to blood pressure would be around 0.02mm, the distance of the skin to the optical fiber would have to fall within this range of 0.25 to 0.55mm, and this would be reflected in the supporting fixture of the device itself during its design. Thus, for a good measure, the distance would ideally be 0.4mm without any surface defects or scarrings.

CHAPTER 5: FABRICATION FOR SEISMIC OPTICAL DEVICE

5.1 FABRICATION PROCEDURES

Before the fabrication of device on an actual SOI wafer can be carried out, test runs were carried out first using normal silicon wafers for familiarization of fabrication process. After successful fabrication on normal silicon wafers, SOI wafers were then used for actual prototype fabrication followed by testing and analysis. SOI wafers used for this device would have 30 μ m device layer thickness, 2 μ m buried oxide layer thickness, and 500 μ m handle layer thickness.

The fabrication is split into five main stages, with the first being the moisture removal bake, followed by photolithography on the device layer (thinner layer). Upon having a layer of photoresist with specific features on the wafer, DRIE will be carried out on the device layer and PR strip will be carried out. After that is done, the wafer would be flipped and photolithography and DRIE would then be done on the handle layer (thicker layer) which requires the wafer to be bonded to another carrier wafer. Subsequently, the buried oxide layer would be removed through HF etching after PR strip is conducted for the handle layer photoresist. Detailed procedures of the process flow are as follows.

For the whole process, one major error that may occur would be in the misalignment after one side of the wafer has been etched as the wafer has to be realigned to the second mask for the second etching step to be done. Rotational and transversal misalignment errors have been studied in APPENDIX C, and it is observed that a maximal tolerance for error would be $\pm 2.5^\circ$ for a 1% chance of error and $\pm 8^\circ$ for a 10% chance of error. This risk is reduced with the use of alignment marks and well-maintained fabrication equipments. Another risk would be in the breaking of the thin structures during each step of the procedure as the wafer would be fragile after etching. Thus, careful handling must be enforced throughout the fabrication while

using proper equipment. A third risk would be in the wafer bonding process, which would be mentioned in the later part of this chapter. Apart from these risks, there are also various parameters in the fabrication process that require attention and investigation. In fact, each step in the fabrication process is critical to the final design of the device itself. A total of 38 steps were required for the fabrication of the devices. From the soft baking to the unique etching recipe of each DRIE process, each parameter would directly affect the resultant layer of photoresist on the wafer and final design of the device itself. These details would be further investigated below, showing the importance of each parameter in each stage of fabrication and Table 11 shows the detailed procedures for the fabrication of the device wafer from the initial bare wafer to the final device.

Table 11. Fabrication Process Flow diagram

Step	Event	Duration	Action	Diagram
1	Wafer Start		Insert wafer
2	Moisture removal bake	20 min	Wafer bake
3	HMDS Priming under vacuum	120 sec	Photolithography on device layer
4	Wafer cooling	5 min		
5	Spin coating of AZ9260 (6μm)	45 sec		
6	Resist Prebaking at 105 degrees	4 min		
7	Resist cooling	5 min		
8	UV exposure	16 sec		
9	Develop using AZ400K	120 sec		
10	DI Rinse	60 sec		
11	Spin dry at 800 rpm	120 sec		
12	Hard bake at 105 degrees	4 min		
13	DRIE on device layer	8 min	DRIE
14	PR Stripping in Acetone	20 min	PR Strip
15	DI Rinse	60 sec		

16	Spin dry at 800 rpm	120 sec		
17	PR Strip in H ₂ SO ₄	23 min		
18	DI Rinse	15 min		
19	Spin dry	120 sec		
20	HMDS Priming under vacuum	120 sec	Photolithography on handle layer
21	Wafer cooling	5 min		
22	Spin coating of AZ9260 (10μm)	45 sec		
23	Resist Prebaking at 105 degrees	4 min		
24	Resist cooling	5 min
25	Mask Alignment and UV exposure	60 sec		
26	Develop using AZ400K	120 sec		
27	DI Rinse	60 sec		
28	Spin dry	120 sec		
29	Hard bake at 105 degrees	4 min		
30	Wafer bonding to other wafer	45 min	DRIE
31	DRIE on handle layer	3 hours		
32	PR Stripping and wafer debonding in Acetone	20 min	PR Strip
33	DI Rinse	60 sec		
34	Spin dry	120 sec		
35	PR Stripping in H ₂ SO ₄	23 min		
36	DI Rinse	15 min	BOE
37	Buffered Oxide Etch using HF to remove BOX layer	30 min		
38	Moisture Removal using supercritical fluid			

5.1.1 MOISTURE REMOVAL BAKE

As the wafers used in fabrication could have been stored for a long period of time, it is always a good practice to heat up the wafer first before any procedure is carried out. This is

done to remove any moisture that would be on the surface of the wafer due to exposure to the environment. This would aid in ensuring almost ideal conditions for the priming process, which would in turn enable better adhesion of the photoresist unto the surface of the wafer. In this case, the wafer would be heated at 105°C for 20 minutes before putting it into the HMDS priming machine.

5.1.2 HMDS PRIMING

Table 12. Original HMDS Priming Recipe

Step	Time	Process
1	30 sec	Vacuum
2	30 sec	Vacuum, N2 purge
3	20 sec	Vacuum
4	10 sec	Vacuum, HMDS
5	30 sec	Vacuum

For the photoresist used in this project, which is the AZ 9260 positive photoresist, HMDS (hexamethyldisilazane) priming is required for good adhesion of the photoresist onto the surface of the Si wafer. HMDS is found to allow a photoresist material to be firmly bound to an oxide substrate adhesively and also helps in preventing an undercutting of an oxide layer during etching.[104] The original recipe used is seen in Table 12 and the results of the initial fabrication are observed in Figure 49. It was found on the first fabricated wafer that the 6um beams for Devices A and B were missing and the beams with bigger widths were warped. Also, the photoresist for the proof mass portions were also found to be washed away during the developing step of the photolithography process. All these suggested that there was poor adhesion of the photoresist on the Si surface during the photolithography procedure.

Figure 49. Adhesion problems found in different device designs

Table 13. Modified HMDS Priming Recipe

Step	Time	Process
1	20 sec	Vacuum
2	10 sec	Vacuum, N2 purge
3	10 sec	Vacuum
4	20 sec	Vacuum, HMDS
5	20 sec	Vacuum
6	10 sec	Vacuum, N2 purge
7	10 sec	Vacuum
8	5 sec	N2 purge

Figure 50. Normal beams and proof mass portions after changing HMDS recipe

A modification of the process recipe was made to allow longer coating of HMDS on the wafer surface. This is done so that there would be better adhesion of the photoresist to the Si wafer surface. The modifications are found in Table 13 and the effects can be seen in Figure 50. The beams and proof mass portions that were missing in the previous iteration were found to be intact on the wafer surface and the beams were straight and not warped. Through this, the importance of proper adhesion of the photoresist layer to the wafer surface can be seen and therefore, the modified recipe would be used for all future wafer fabrication for this project.

5.1.3 SPIN COATING

Spin coating is a process whereby a thin layer of photoresist is coated on the surface of the Si wafer. A wafer is placed on a vacuum chuck in the middle of the spin coater machine with an amount of AZ 9260 photoresist subsequently deposited manually onto the wafer before

a program is initiated to start the spinning of the vacuum chuck and wafer. The photoresist is then uniformly spread throughout the wafer surface and subsequently thinned due to the continual spinning of the wafer at a certain speed. The thickness of the photoresist layer is determined by the spinning speed, ramping rate and duration of spin. Also, for each different procedure, the requirement for the thickness of the photoresist is determined by multiple factors, like the smallest feature size, etch depth and aspect ratio. Standard process parameters for the CEE Spin Coater machine in the Micromachines Lab 1 are given in Table 14 below.

As a requirement to determine the thickness of the photoresist, the photoresist coated must be able to last through the whole duration of the DRIE process, while at the same time, the photoresist thickness must be able to produce the 6 μ m beams accurately. For the device layer etch, the thickness of the photoresist is determined to be within the range of 5.5 μ m to 5.8 μ m after trial and error, which would be the Standard Program 5 in the Spin Coater system. As for the handle layer etch, the thickness of the photoresist is to be within the range of 9.5 μ m to 10.7 μ m, which is Standard Program 2.

Table 14. Standard programs for CEE Spin Coater

Standard Program	2	3	4	5
Target Speed (rpm)	2000	3000	4000	5000
Spin Duration (rpm)	30	30	30	30
Thickness Achievable (um)	9.5 - 10.7	7.1 - 7.4	6.0 - 6.3	5.5 - 5.8
Dispense Type	Manual			
Vel/0 -To spread resist and cover whole surface (rpm)	500	500	500	500
RMP/0 -Ramping (rpm)	100	100	100	100
Time/0 (sec)	10	10	10	10
Vel/1 -Actual coating (rpm)	2000	3000	4000	5000
RMP/1 -Ramping (rpm)	5000	5000	5000	5000
Time/1 (sec)	30	30	30	30

5.1.4 SOFT BAKE

Soft baking of the wafer coated with photoresist is needed to remove the remaining solvent concentration of the photoresist. This process is required to avoid mask contamination, improve resist adhesion to substrate, prevent popping or foaming of N₂ gas created during UV exposure, reduce dark erosion during development and also prevent bubbling during subsequent thermal processes carried out on the wafer itself. [105]

The use of heating equipment for soft bake is also very critical. Hot plates are strongly encouraged over using of ovens as they employ direct application of heat through conduction, which would enable faster and more uniform heating of the entire wafer. Also, hot plates conduct heat through the wafer unto the photoresist layer, which allows the purging of gases from the internal layer to the external surface of the thick film of photoresist. If an oven is used, the thermal transfer is through convection currents unto the external surface of the photoresist. This heating would cause the external surface of the photoresist to be heated up first which can subsequently cause trapping of gases in the photoresist layer after time. This would result in poor photoresist resolution and also bubbling in the photoresist layer during exposure.

For baking parameters, the two variables would be temperature and time. If the temperature is too low or exposure timing too little, bubbles will be trapped in the photoresist layer that can only be noticed after hard baking, which would be more difficult to remove. If the temperature is too high or time taken is too long, the rate of development would decrease and also cause the photoresist to become brittle. Therefore a good measure for the soft bake would be 105 degrees at 4 minutes using a hot plate for the device layer. As the photoresist layer is thicker for the handle layer, the soft bake timing would increase to 8 minutes while still at 105 degrees.

5.1.5 UV EXPOSURE

Photolithography employs the use of UV light to remove the exposed areas of the positive photoresist AZ 9260 through the use of a glass mask. By doing so, the aim is to create a layer of photoresist with the exact pattern found on the mask. This layer of photoresist is then used as a mask for the etching process that follows after photolithography. Similar to the baking parameters mentioned above, there is only a small time window for good exposure. If the exposure time is too short, the UV light would not be able to penetrate the full thickness of the photoresist. This would result in difficulty during the developing of the photoresist as there still is a thin layer of photoresist remaining on the exposed areas. If the exposure time is too long, the UV light will be reflected into the surrounding regions which are not supposed to be exposed. These affected regions would then also be cured and washed away with the exposed photoresist, resulting in inaccurate features being created. In worse case scenarios, small features would also be removed.

The main parameters for the photolithography process include the exposure time and exposure power used for the UV lamp. The exposure power cannot be controlled from the user end of the equipment, thus the default setting of 11.5 Watts was used throughout all photolithography iterations. On the other hand, the exposure timing for the process can be

controlled by users. For the exposure of a 6 μ m thick layer of photoresist, 12.5 seconds of UV exposure is deemed to be appropriate with good features. As for the exposure for the 10 μ m thick photoresist for the handle layer, a multiple step exposure of 6 sets of 10 seconds is used. This multiple step exposure is to ensure that the energy from the UV lamp penetrates the photoresist layer all the way to the bottom, where the wafer is with reduced thermal effects. If the exposure was a single set of 60 seconds, the penetration of the energy would result in thermal effects on the surrounding photoresist.

5.1.6 DEVELOPMENT

After photolithography, the wafer is placed in a container filled with 75ml of developer mix that is diluted with deionized water in the ratio of 1:2 (AZ 400K developer: deionized water). The mixture will gradually remove the UV exposed positive photoresist through a chemical reaction, exposing the Si wafer surface at specific areas so that etching can be carried out on the wafer surface. Although this, the developer mix will also slowly remove the unexposed photoresist and this is generally termed “dark erosion”. This means that the wafer cannot be exposed to the developer for long periods as it would eventually also remove all the unexposed photoresist, reducing the accuracy of the features. In this case, the main parameter for the developing stage would be the time taken for developing. With regards to the concentration of developer used, it is based on the standard composition given in the Micromachines Lab 1, thus it is advisable to not change the concentration unless required. A good example of how a wafer looks like under a microscope after development is the left image of Figure 51, where the lighter portion is the bare silicon surface and the darker regions are the photoresist layer.

5.1.7 HARD BAKE

Before sending the wafer for etching, there must first be a baking procedure to reduce the remaining solvent content found in the photoresist layer that's left on the Si wafer. During this process, the photoresist layer will also be strengthened, however, at high temperatures, the resist will start to reflow, which would affect the design of the device. For the AZ 9260 photoresist, the resist starts to experience rounding of features at temperatures approximately above 110°C. For this procedure, the parameters would be the temperature and time taken. Since rounding starts at 110°C, the temperature chosen would be close to that but not above it, which would be approximately 105°C. The time taken for this hard baking should not be too long as the longer the wafer was heated, the more reflow and rounding of features would occur.

Figure 51. Rounding effects of hard baking can be seen when comparing the original (left) and resultant (right)

Table 15. Effects of hard baking on dimensions

Hard baking	Critical variable	Mean	Target	% Diff
Before	Gap width	27.08	30	-9.72
	Beam width	9.76	7	39.37
After 10min @ 105 deg	Gap width	27.56	30	-8.14
	Beam width	9.63	7	37.53

Thus, as a good measure, the time taken for this process would be 10 minutes. The effects of hard baking at this temperature and timing can be seen in Figure 51. A comparison table of the effects of hard baking is given in Table 15 where it can be observed that the main effect of hard baking is seen in the rounding of the features while the dimensions remain relatively the same. This rounding of the edges however, would affect the etching stage as the thickness of the photoresist is not uniform, which may lead to over etching of critical structures.

5.1.8 ETCHING

Figure 52. Schematic of DRIE a) Sidewall passivation using C_4F_8 , b) Isotropic etching using SF_6 gas c) SEM image of deep trenches d) Magnified SEM image of sidewall with scallops. [106]

For the etching process, a Deep Reactive Ion Etching process is used. This process is a dry process that allows highly anisotropic etching that can be used to create deep trenches and holes in Si wafers. There are two distinct steps in this process, the etch step and passivation step, where different gases are used. Figure 52 shows a good illustration of how the process works.

For the equipment in the Micromachines Lab 1, the parameters that can be modified would be the total number of cycles and the time taken for each etch and passivation step (See Table 16). The gas ratio recipe for the machine is also given in Table 17. As mentioned above, there would be two different DRIE steps, a 30um etch for the device layer and a 500um etch for the handle layer. Therefore, there needs to be two different recipes for the two different depths. Also, in consideration of their different difficulties in etching, the device layer would be etched using the machine in Micromachines Lab 1, whereas the handle layer etch would be done in the School of Electrical & Electronic Engineering where there is another DRIE machine with higher control for deep etching. The DRIE recipe for that machine is given in Table 18 and Table 19. For this handle layer etch to be done, the 4” Si wafer has to be bonded to a 6” wafer before etching can be done as the deep etch is almost through wafer and would damage the machine if not done. After each etching stage, PR strip has to be done to remove the remaining photoresist layer on the wafer.

Table 16. Modified DRIE Recipe for Device Layer Etch

General	Value	Units
Process Time	0.08:34	hh:mm:ss
Parameter Switching Cycle Times		
Etch (On Time)	8	ss
Passivate (On Time)	8	ss
Total Cycles	32	

Table 17. DRIE Gas Ratio Recipe for Device Layer Etch

		Etch	Passivation
Line	Gas Name	Flow (sccm)	Flow (sccm)
1	C4F8	0	100
2	SF6	130	0
3	O2	13	0
4	Ar	0	0
5	CF4	0	

Table 18. Modified DRIE recipe for Handle Layer Etch

General	Value	Units
Process Time	03:00:00	hh:mm:ss
Parameter Switching Cycle Times		
Etch (On Time)	10	ss
Passivate (On Time)	3	ss
Total Cycles	762	

Table 19. Gas recipe for DRIE for Handle Layer Etch

		Etch	Passivation
Line	Gas Name	Flow (sccm)	Flow (sccm)
1	C4F8	0	85
2	SF6	200	0
3	O2	0	0
4	Ar	0	0
5	CF4	0	

5.1.9 PR STRIP

After the etching phase, there are two different methods of PR stripping employed. The first is using acetone for approximately 20 minutes, where the remaining bulk AZ9260 photoresist is removed from the wafer, and next would be using sulphuric acid. This is because even though the bulk photoresist is removed, a layer of residue remains on the surface of the silicon wafer. This layer has to be removed as the layer is inconsistent and would cause interference in the optical path between the LED and the phototransistor. Also, it will affect the heat transfer within the wafer bonding between the 4" and 6" wafer during the handle layer DRIE. To do so, a wet process was utilized employing the use of sulphuric acid (H_2SO_4) in the wet bench. The wafer was placed in a bath of H_2SO_4 and left for 23 minutes at a temperature of 120°C . It is then removed and placed carefully into a bath of deionized water and left for 5

Figure 53. Optical images of devices before PR strip in sulphuric acid (a & b), and after PR strip in sulphuric acid (c & d)

minutes to wash off the acid remaining on the wafer. At this point of time, the wafer is relatively brittle, thus instead of the normal process of spraying the wafer with deionized water, baths of water are used instead. It is subsequently placed into another bath of DI water two more times before drying to ensure that the sulphuric acid is completely removed from the wafer. To dry the wafer, the spin dryer or nitrogen gun cannot be used. Thus, the wafer is to be placed in a supercritical CO₂ fluid and boiled to remove the moisture on the wafer while at the same time preventing stiction to happen to the thin beams of the structure.

As observed from the optical images in Figure 53, the residue from the DRIE process and wafer bonding process is quite opaque. Effects from the layer would be very obvious if it was not treated. This residual resist layer would also reduce the effectiveness of the device in terms of the mobility of the proof mass during operation as the resist layer was covering the proof mass and frame at the same time. Also, by removing this layer in between the device layer etch and wafer bonding step between the 4" to the 6" carrier wafer, it helps with the bonding and subsequent handle layer etch. This is because the thermal dissipation through the carrier wafer during the etching phase would be hindered by the thin residual layer if not removed.

5.1.10 WAFER BONDING

Wafer bonding is needed before the handle layer etching to protect the DRIE machine from damage during the almost through etch procedure. The bonding process makes use of the same photoresist as a bonding matrix to stick the two wafers together. This process is also a critical component for the handle layer etch as poor bonding could result in a few problems in the outcome of the etching procedure. One scenario would be if the two wafers were not placed completely flat against each other and pressed together, there would be an angular difference and during DRIE, the etching would be faster on the side that is further away from the 6" wafer due to heat buildup that would speed up etching. Because of this, the top layer of photoresist

that is on the top surface of the 4" wafer might be removed faster than expected, exposing the bare wafer surface to ionic bombardment during DRIE. Another scenario would be when there are air bubbles trapped in the bonding matrix when the 4" wafer is placed on the bonding matrix. The presence of these air bubbles would cause thermal non-uniformity on the wafer surface, leading to a similar situation as the scenario mentioned above.

The parameters would be the thickness of photoresist used for bonding the two wafers and the time used for heating the wafers during the bonding process. If the photoresist layer is too thick, there would be poor thermal conduction through the photoresist layer, which would cause poor thermal uniformity during heating processes and also etch inaccuracy during the subsequent DRIE procedure. If the photoresist layer is too thin, there would be very little bonding between the two wafers, which can result in debonding even before the DRIE procedure can be started. Therefore, as a good standard, a 16µm thick layer of AZ 9260 photoresist would be coated on the 6" wafer in a two-stage process. An 8µm thick layer is first coated and hard baked at 110°C for 140sec, followed by another 8µm layer of AZ 9260. The 4" wafer is then placed in the center of the wafer manually, pressed down and baked again at 120°C for 10 minutes again to bond the two wafers in place. Following that, the two bonded wafers are then hard baked at another 120°C again for 10 minutes again before being sent for handle layer DRIE. The final results of the fabrication are seen in the next chapter.

5.1.11 EFFECTS OF DEVICE LAYER ETCHING ON HANDLE LAYER

ETCHING

Before the integration of etching for both layers, optimization for the handle layer DRIE was carried out and the results were good throughout the wafer (See Figure 54). The grating walls had good profiles and the etch rate was relatively constant throughout the whole

Figure 54. Handle Layer DRIE before integration with Device Layer DRIE showing good grating width uniformity

wafer. However, upon integration with the device layer DRIE, multiple issues surfaced like the bowling effect on the grating walls as observed in the green highlight of Figure 55, the inconsistent dimensions for the grating and gaps as highlighted orange, and the undercutting issue in the grating as seen in the red box.

These different issues were mainly due to the temperature management problems that occurred within the bonding matrix as the photoresist layer in between the 4" and 6" wafer was not uniformly distributed. This is because the surface of the 4" wafer that was in contact with the bonding matrix was already etched previously. This mean that there was no proper contact with the bonding matrix as there would be pockets of air within the surface itself. With the

pockets of air, thermal distribution of the handle layer surface of the wafer would be very poor, leading to differing etch rates throughout the wafer itself.

This problem is a critical issue for the fabrication of the device and modifications had to be carried out on the fabrication procedures to rectify this issue. One modification made was the wafer bonding portion, where the 16 μm thick photoresist had to be deposited. Instead of spin coating the bare silicon wafer surface twice with the AZ 9260, only 8 μm was spin coated unto the surface. The other SOI wafer surface where the device layer was already etched was then coated with the other set of 8 μm thick photoresist before being combined together to form the 16 μm thick photoresist layer for wafer bonding. This method was first tested out then subsequently used for the final fabrication as the results produced were successful, and devices were developed that could be used for further testing.

Figure 55. Effects of integration of etches for both layers

5.2 RESULTS OF FABRICATION

Upon completion of all the steps given above, the wafers were outsourced for fabrication to Nanostructures Inc. to speed up the fabrication of wafers as it was very time-consuming to be carried out in the cleanroom. A full view of the wafer is given in Figure 56, which shows the remaining working devices after checks and experiments have been conducted on the wafer. Many of the devices were lost during the transferring of the wafer during experiments and also during the transportation stage before the wafer was received. The results of the fabrication can be seen under the microscope in Table 20 and SEM images to show the results of the DRIE procedures were captured in Figure 57.

Figure 56. Full wafer showing remaining successful devices

Table 20. Microscope images of succesful devices

Device Layer View (Top view)	Handle Layer View (Bottom view)
Device A

Device B

Device C

Device D

Device E

Device F

Device G

Device H

Figure 57. Scanning electron microscopy images of the final product with integration of the gratings after through wafer etching (left) and the device layer etching (right)

Various difficulties were encountered in the fabrication procedures. As the handle layer DRIE is a 500 μm deep etching, the etching of the side walls of the gratings were unable to be uniformly etched through, causing underetching near the corners of each grating. This in turn caused the rectangular gratings to be elliptical as observed in each of the devices in Table 20. In the left image for Figure 57, each column is observed to be thicker at the base in a non-uniform manner. This thicker base is due to the underetching during the handle layer DRIE, and the non-uniformity is because the proof mass was broken non-uniformly, causing uneven breaking points in each column. Despite this, each column was still observed to be identical to each other within the same proof mass. The underetching would result in less light entering the photodetector beneath, however the experiments have shown results that were still comparative with other devices developed using this same method, thus it was still acceptable for use.

CHAPTER 6: EXPERIMENTAL METHODS & RESULTS FOR SEISMIC MOEMS DEVICE

With the simulated results, the optimized device designs were sent for fabrication using the wafer design given in Section 4.1.4. Subsequently, upon receiving the device designs, experiments were conducted to determine if the devices are able to detect any movement upon external forces acting on it. However, before the MOEMS layer was received, a large prototype was first developed for preliminary experimentation. Subsequently, when the MOEMS wafer was received, two different photodiodes were used to test the device due to the different outputs both devices provided. All three experiments were discussed in this chapter.

6.1 EXPERIMENTATION USING PROTOTYPE

Figure 58. Experimental setup for mock up

Figure 59. Image of the stainless steel mock up (left) and zoom in of beam (right)

Figure 60. Results obtained from experiments on mock up

Figure 61. Close up of results show a distinct pattern in the waveform when force applied is in different directions

The aim of this first experiment is to confirm whether the device concept would be able to detect the motion introduced to it. An enlarged version of the proof mass layer has been fabricated using stainless steel based on Device Design D to observe the effects of motion on the device design (See Figure 59). This mid scaled proof of concept is used in the experimental setup as seen in Figure 58. The proof mass was clamped directly above a PD300 photodiode power sensor with a laser module placed 45cm above the proof mass. The distance was determined after considering the beam divergence angle with the coverage of the width of the proof mass, ensuring that the proof mass gratings had a uniform exposure. The movable sliding rail was then given a series of pushes to show if there would be any change in the readings captured by the photodiode below. The pushes were delivered by hand as at this point of time as the linear transducer that was originally planned for use was not functioning and have to be sent for repairs.

Based on the results obtained in Figure 60, it was observed that there were obvious increases in the light intensity upon every exertion of force on the setup. This means that the device was able to detect its motion, which was at the instant the force was exerted. A zoom in of each waveform shows a distinct pattern when force is applied in opposite directions (See Figure 61). Each of the curves magnified in Figure 61 are marked according to their colors in Figure 60. Both curves show a ringing in their waveforms, which shows that the system is underdamped. With reference to the arrow direction given in Figure 58 as the forward direction, when the force is in the forward direction, there would be a dip in the waveform followed by a peak. When force is in the opposite direction, it would be the other way, with the peak first before the trough. This can be used as a reference for future experiments.

6.2 EXPERIMENTATION USING POWER SENSOR

The aim of the second experiment was to observe the effects of the acceleration on the device readout. The experimental setup was developed as according to the setup shown in

Figure 62. It consists of an 850nm IR Laser Module (MM850-5, US-Lasers Inc.), a photodiode power sensor (S121C, Thorlabs) and power meter console (PM100D, ThorLabs), and oscilloscope (MSO1104Z, Rigol). The power meter console is subsequently connected to a laptop for direct data capturing. A reference accelerometer (Miniature Single Axis Piezoelectric Accelerometer Model No. 353B15, PCB Piezotronics) was used for calibration and comparison of results. The accelerometer was connected to a signal amplifier, which was connected to an oscilloscope for data capture and storage.

The fabricated wafer is fixed unto a customized wafer holder, which is placed on a pair of sliding rails and the laser module is placed perpendicularly on top of the test device with the power sensor aligned directly beneath it, as shown in Figure 62b. The sliding rails were placed

Figure 62. (a) Schematic diagram of experimental setup (b) Cross sectional view of wafer on wafer holder showing alignment of laser module and power sensor.

Figure 63. (Left) Motor used for force exertion. (Right) Schematic diagram that shows crank angle used in calculations.

parallel to each other and manually adjusted to be parallel with the surface of the vibration isolated table. To move the wafer holder, a gearbox (Tamiya 72008 4-Speed Worm Gearbox) which is able to produce a torque of 2,306 gf.cm, generating a force of 10.42N at maximum according to specifications was used (See Figure 63). The motor moves in a rotary motion, similar to a single piston motor based on the equations[107]

$$X = R\cos\theta + \sqrt{L^2 - R^2\sin^2\theta} \quad (1)$$

$$X' = -R\sin\theta - \frac{R^2\sin\theta\cos\theta}{\sqrt{L^2 - R^2\sin^2\theta}} \quad (2)$$

$$X'' = -R\cos\theta - \frac{R^2(\cos^2\theta - \sin^2\theta)}{\sqrt{L^2 - R^2\sin^2\theta}} - \frac{R^4(\cos^2\theta\sin^2\theta)}{\sqrt{L^2 - R^2\sin^2\theta}^3} \quad (3)$$

Upon activation of the motor, forces are induced on the wafer in a linear direction due to the guiding rails placed below it.

All experiments were conducted in a dark environment as the ambient light produces an effect on the output detected by the device photodiode. Throughout the experiment, no linear amplifier was used to amplify the output of device. An inherent disadvantage of this experimental setup would be the weight of the wafer holder with the photodiode and laser module, which is approximately 200g. This reduces the acceleration experienced by the device due to the friction on the guide rails and styrene guides attached to the holder. The coefficient

of friction between these two materials is approximately 0.325, resulting in a frictional force of 65mN.[108] The wafer was placed on the wafer holder and the device design which had the most effective results is shown below, which in this case is Device E. The dimensions for the proof mass parameters are given as follows.

Device Layer Thickness:	30 μm
Handle Layer Thickness:	500 μm
Beam Length (B_L):	210 μm
Beam Width (B_W):	6 μm
Gap Width (G_W):	30 μm
Grating Length (G_L):	1500 μm
Mass Area ($M_W \times M_L$):	2 mm x 1.8 mm

In the experiments, it was standardized that the first 50 seconds was to measure the baseline whereby no force was introduced. The measured intensity at that point is average at 558 μW . After that, the motor was turned on from 50 seconds to 140 seconds. The sensor showed irregularities when the detected values were below the baseline previously mentioned. This is because the motor experienced some difficulty in engaging the crank arm when the arm was fully retracted, resulting in additional variability that will not be characterized here. This additional variability resulted in fluctuations in the readout below the baseline. After turning on the motor, it exerts a periodic force on the device as shown in Figure 64. A series of peaks and troughs were observed with each periodic motion, with the first period different from the others as the motor was just turned on. The noise floor of the device is based on the first 50 seconds where an average of 558 μW was detected.

Figure 64. Experimental observations obtained from device at quiescent state with motor excitation.

Figure 65. Theoretical evaluation of piston motion

Figure 66. Mapping of peaks from experimental measurement to acceleration

As the device readout was affected by the motor when the readout was below the baseline, no information was able to be obtained from it. Each period, except the first, was extracted and mapped onto a single waveform to show the average value and error bar at each point (See Figure 65). From this, it can be seen that there is low standard deviation at the peak of each period with the average value at $563\mu\text{W}$. The correlation between the forces exerted on the device and the detected intensity of light by the photodetector is given in, where a sensitivity of $2.5\mu\text{W/G}$ is calculated from the results (See Figure 66).

6.3 EXPERIMENTATION USING FPS-1 PHOTODIODE

The previous experiment showed that the intensity levels during the movements were substantial and detectable, however, the detection rate of the photodiode power sensor itself is not fast enough. Therefore, a new set of experimentation had to be carried out using a faster detection. The aim of this new experiment is to directly compare the readout obtained from the device with the readout from a commercial accelerometer. The new experimental setup makes use the same piezoresistive accelerometer, laser module and fixtures, as given in Figure 67. This accelerometer was checked to produce 158mV for acceleration of 10m/s^2 at rms value or

10 μ m displacement at rms value. Based on the simulated model, the proof mass would produce a voltage of 2.07mV upon an acceleration of 2G. An FPS-1 photodiode purchased from Ophir Photonics and subsequently, both the commercial accelerometer and photodiode are connected to a digital oscilloscope for data capturing. Another modification to the experimental setup was

Figure 67. (a) Schematic diagram of experimental setup (b) Cross section of wafer showing position of laser module and photodiode aligned with proof mass.

Figure 68. Experimental observations obtained by MOEMS device and commercial accelerometer undergoing similar forces from rotary motor.

a set of gratings which is placed on top of the wafer, which would subsequently be pressed down by the laser module held perpendicularly over it. The gratings would be used for differential measurement as the photodiode used is not the type of photodiode designed for the device as given in Figure 27.

Experimental observations where a series of revolutions of the rotary motor is applied on the MEMS device wafer is given in Figure 68. From the graph, the device readout successfully captured the peaks and troughs similar to the commercial accelerometer waveform. Although the resolution of the device is incomparable to the resolution of the commercial accelerometer, it is still able to show the exact positions of the highest acceleration experienced by the commercial accelerometer. A further comparison of the waveforms with respect to the calculated acceleration experienced by the proof mass is given in Figure 69. A linear relationship with gradient of 0.318 was observed in Figure 69a when comparing the different methods of detection. The results were tabulated over eight periods within each run. The device readout was then compared against the acceleration experienced by the proof mass. A trendline of $y = 0.0136x + 0.0135$ was observed. Based on the trendline, the sensitivity of the optical device was 13.6mV/G and the estimated noise level is 13.5mV. This high noise level is due to the unavoidable friction between the guiding rails and wafer holder, coupled with vibration noise of the motor. The error bars given in Figure 9b are large as the motor used was made using gears that are easily worn out. On top of that, the alignment of the photodiode, proof mass and laser module during each experiment run also affects the accuracy.

Figure 69. (a) Comparison of device readout against commercial accelerometer readout.

(b) Comparison of device readout against calculated force.

6.4 RESULTS AND DISCUSSION

Based on the results obtained from all the above experiments, the device has shown a sensitivity of 44.103mV/G with noise level at 13.5mV . To relate it back as a seismic detection sensor, a signal would be detected when there is a source providing an axial force along the direction of the proof mass. Through the comparison of the loading of each foot from [27] with

the results obtained in the experimental setups, it is observed that the device mentioned in this project would be able to detect people walking on a light weight floor structure as mentioned in the paper. It is also possible to say that seismic motion occurred from heavier and louder sources like troop movements can also be detected by the device as these motions are generally heavier than regular walking.

Given that the footstep amplitude responses are not very high generally, and that the amplitude response gradually decreases the further it is from the source, the device developed can be used for either indoor seismic motion monitoring or close proximity outdoor motion monitoring near fences or gates. Also, the overall sensor itself is small, thus it can also be used by spreading across wide spans to improve detection coverage.

Currently, the footstep amplitude response has been taken from literature sources due to certain limitations of the project. Also, given that the experimental setup was improvised, it is unavoidable that the minimum detectable acceleration is currently at this level. With better facilities and equipment that can be used, it is believed that the minimum detectable acceleration would be even lower than what it currently is now as the current experimental setup would inherently have large friction to overcome on the guide rails. The electronic circuitry for the sensor has also been designed and is ready for fabrication for integration with the device for experimental integration (See APPENDIX F). The design is based off a similar design by Hortschitz and thus it is proven to be work upon integration with the device layer.

CHAPTER 7: EXPERIMENTAL METHODS & RESULTS FOR WEARABLE OPTICAL BLOOD PRESSURE SENSOR

Experiments were only able to be conducted for Device A as the optical waveguide was easily obtainable and could be fabricated in house. The ball shaped side-fire waveguide were unable to be purchased for experiments due to various reasons and thus will not be mentioned in this chapter.

7.1 EXPERIMENTATION FOR DEVICE A PROTOTYPE

Figure 70. Setup used for evaluation of light intensity against displacement

The first experiment conducted is based on the optimized curved waveguide dimensions that were obtained from simulations, and the designed waveguide was experimented on using the experimental setup as seen in Figure 70. In the designing of this device, for the device to be able to detect and give accurate blood pressure values, two of this optic waveguide devices would be needed and concurrently used. However, the optic waveguide device has to first proof that it is able to effectively obtain the blood pressure waveform, thus this series of experiments using only one setup. A comparison was done with another rectangular waveguide that has a 45 degree chamfer on the corners of the edges so that the light path can be reflected within the optical waveguide itself. The setup makes use of the same 850nm laser module MM8505I from

Figure 71. Simulation results showing the comparison between polished and unpolished curved and rectangular waveguides (See inset).

US-Lasers as the light source and an evaluation board ISL29035 that has a photodiode power sensor for the detector. From the graph in Figure 71, it is found that the polished curved waveguide shows the most change from the 0.3mm to 0.9mm displacement. Using this set of results, the waveguide is then used for a direct skin contact test of the device itself. As the setup for the direct implementation does not have space for the evaluation board, another photodetector had to be used. For this, a PD300 photodiode intensity sensor attached to a Nova interface from Ophir Photonics was used instead, as seen in Figure 72.

Figure 72. Experimental setup used for detection of blood pressure from radial artery.

Figure 73. Results of initial experiment with optical waveguide.

The results obtained from the experiment shows significant peaks at regular intervals of the waveform (See Figure 73), however the baseline intensity was erratic, thus making the findings inconclusive. With this set of results, the pulse rate of a human cannot be correctly determined although there are peaks observed at regular intervals of 0.9 seconds, equivalent to a pulse rate of 67 beats per minute that is within the range stated for humans within the age of 21-45.[109] A low pass filter was considered however the equipment provided did not have the function, therefore another sensor had to be used. Apart from this, there is also a need to

use a pulse transducer as a reference to determine the accuracy of the device. As long as the pulse rate can be determined, the next step would be the use of two devices to determine the blood pressure values.

7.2 EXPERIMENT FOR TIME LAG BETWEEN DEVICE AND PULSE TRANSDUCER

As different sensors are to be used in the subsequent experiment, there could be a time lag in the capturing of the readings from both the pulse transducer and the device, thus experiments have to be carried out to determine the magnitude of such an error as it could be a

Figure 74. Experiment to test time lag between the setup for the optical waveguide device and pulse transducer

large source of error for future experiments. This experiment was done by placing the pulse transducer right beneath the movable part of the waveguide and gently tapping the waveguide. The results were given in Figure 74, showing an average time difference of 0.055 seconds with standard deviation of 0.0279 seconds. This would mean to say that if there was any other difference in timing during subsequent experiments, it would be due to the pulse transit timing

(PTT), which is the target variable when two sensors are used. This experiment raises an issue of the optical waveguide not being able to return back to its original state effectively, given that the device readout did not go back to the original values after tapping. However, looking on the bright side, it was observed that the device could have very good readings even when the device was only gently touching the skin surface, meaning that there was no need for any cuffs or bands to be used for this device and a simple adhesive would be good enough for attachment to the skin surface.

7.3 EXPERIMENT USING DIRECT IMPLEMENTATION

Figure 75. Experimental setup showing placement of pulse transducer and optical waveguide for device

For an improvement on the first experiment, the same mock up was used together with a pulse transducer using a LabChart software according to the setup found in Figure 75. The pulse transducer is placed on the tip off the middle finger and not on the exact same position as the waveguide itself as preferred but the main components of the pressure waveform, like the pulse rate, timing of the DBP and SBP can still be inferred. The time lag between the two detection mechanisms obtained in the previous experiments were eliminated from the raw data before the graph was plotted too.

Figure 76. Experimental observations recorded from pulse transducer (blue) and optical device (red).

Based on the graph for the device (See Figure 76), a periodic waveform can be captured by both the pulse transducer and device, where there is a consistent increase and decrease in the peaks of the pulse transducer waveforms and overall waveform of the device. This pattern was found to have the same period as the respiration rate of the human subject, who in this case would be myself, of approximately 12.35Hz. One thing to note would be that the curves for the pulse transducer and the device are inverted, which is why the secondary vertical axis is inverted. This is because the mechanism of the optical device, whereby an increase in displacement of the waveguide due to increase in blood pressure, results in the deflection of the movable arm away from the optimal position in line with the photodetector.

The peaks of the pulse transducer are representative of the systolic blood pressure and the frequency of this is 1.188Hz. Likewise, the peaks observed by the optical device are also of a similar frequency of 1.99Hz.

7.4 RESULTS AND DISCUSSION

Using the results obtained from the latest experiments, it is found that the device was able to capture the pulse rate of a human through the evaluation of alternate peaks in the device waveform. This is because the device is sensitive enough to capture the secondary peaks in a typical radial artery waveform. Also, respiration effects are also felt in the waveforms. With the usage of two of the devices, the comparison of the pulse transit timings (PTT) would ignore

Figure 77. Focused comparison of experimental observations (Top) with ECG observations found in [110] (Bottom)

the effects of respiration and also motion artifacts. Also, the locations of the two devices can be as according to what were used in the experiments, which has been proven to be able to detect pulse waveforms effectively. With the use of two devices for the detection, it is possible to determine the blood pressure values based on the PWV theory given in Chapter 3, however

for the current experiment, only one device was used so no comparisons could be made. However, using the information obtained based on the experimental results, the PWV calculated is estimated to be 7.929m/s. A reference value for a males within the age of 26-41 is mentioned by Koivistoinen et al [111] to be an average of 7.7 m/s with a standard deviation of 1.2m/s. By using the equation given in Chapter 3 for the PWV,

$$PWV = \sqrt{\frac{Eh}{2r\rho}} = \sqrt{\frac{E_0 e^{\alpha P} h}{2r\rho}}$$

the systolic blood pressure can thus be obtained after initial determination of user-specific variables like E_0 , α , r , ρ and h . After the systolic blood pressure is determined, two points of the performance curve, the SBP point and 0 point, can then be used to determine the estimated diastolic blood pressure of the user based on the DBP point on the curve during detection. Also, as this values are for the peripheral blood arteries, the GTF function mentioned in Chapter 2 can then be used to obtain the values of the central blood pressure (See Figure 16).

One additional piece of information that is interesting to note would be that the pulse transducer waveform obtained in the experiments was similar to the conventional waveform obtained by an electrocardiogram (ECG) as observable in Figure 77. Multiple repetitions of the experiment were carried out but the waveform obtained still had similar characteristics. This observation could show a relationship between finger pulse waveforms and ECGs, where further investigations may lead to further breakthroughs in the medical industry.

To conclude, experiments has confirmed the ability to detect pulse rates from the radial artery using a single device from Design A, which is the initial step to the detection of blood pressure. Measurement results however have observed high motion artifact due to respiration, which would be difficult to be removed using only one device. With this, it is believed that Device C would also be likely to have such effects too. As a result, it is believed that designs that employ two devices would be more reliable and effective in detecting blood pressure.

CHAPTER 8: CONCLUSION AND FUTURE WORKS

In this project, two different approaches in optical motion sensors have been discussed. One method makes use of MEMS fabrication technology together with optical gratings for seismic detection that can be used for security and defense purposes. Another method employs the use of optical fiber or waveguides for the detection of human blood pressure with the aim of making the sensor wearable. Research studies have shown that the seismic MOEMS device concept developed in this project has a strength in terms of being one of the devices that have higher resonant frequency with better sensitivity. Although the output is a lot less as compared to other state of the art accelerometers, the developed device has shown have a device sensitivity of 44.103mV/G. The experiments conducted show the effectiveness of this design concept in capturing motion and the waveform generated by the device was the same as the commercial accelerometer used as a control in the experiments. This device also has the novelty of integrating an optical MEMS sensor for use in seismic motion sensing. The fabrication of the device has also displayed the capabilities of a MOEMS device making use of double-sided DRIE using SOI wafer. Future work for this device would involve the integration with a CMOS photodetector and experiments on actual human footstep detection.

In the second part of the project, with regards to the wearable blood pressure sensor, all design approaches were simulated using optical simulations however due to limitations of the project, only the first design concept was able to be fabricated and experimented on. Albeit this, theoretical formulation were still found to link the pulse wave velocity values obtained to pressure values. Experimental values obtained were conducted on the PMMA waveguide device and the results obtained showed that it was successful in obtaining the pulse rate when a single device was used together with a pulse transducer. However the data captured also showed the effects of respiration and also possibly effects of residual strain and stresses causing the detector readings to fluctuate. For the other two design approaches, they would be attractive

and reliable given that a novel use of the ball-shaped side-fire optical fiber tip is used to allow devices to lie along the surface of the skin, unlike other detectors, and also to be cuffless. If the other two device approaches were to be fabricated, they would be able to successfully detect blood pressure as observed through simulations. Cuffless blood pressure sensors are still being explored and the design approaches mentioned here would also be of huge impact in the wearable devices arena. Thus the next step in this approach would be the fabrication of the optical fiber tip and experimenting on a human subject.

In conclusion, all deliverables of the project have been met and the innovation of these two devices would be of huge benefit in bringing cutting edge aspects of research in MEMS and optics fields closer to consumers, while at the same time bringing innovation into the aspects of optical motion sensing and optical wearable technology.

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APPENDIX A

Table 21 :Seismic Accelerometer Comparisons

Author	PCB Piezotronics 393A03
	Sensitivity	1 V/g
	Resolution	0.0001 g rms
	Resonant Freq	≥ 10 kHz
	Noise	-
	Freq Range	0.3 to 4000 Hz
	Method	Piezoelectric
Author	PCB Piezotronics 393B04
	Sensitivity	1 V/g
	Resolution	0.000003 g rms
	Resonant Freq	≥ 2.5 kHz
	Noise	-
	Freq Range	0.05 to 750 Hz
	Method	Piezoelectric
Author	PCB Piezotronics 393B05
	Sensitivity	10 V/g
	Resolution	0.000004 g rms
	Resonant Freq	≥ 2.5 kHz
	Noise	-
	Freq Range	0.5 to 750 Hz
	Method	Piezoelectric

Author	PCB Piezotronics 393B12
	Sensitivity	10 V/g
	Resolution	0.000008 g rms
	Resonant Freq	≥ 10 kHz
	Noise	-
	Freq Range	0.1 to 2000 Hz
	Method	Piezoelectric
Author	PCB Piezotronics 393B31
	Sensitivity	10 V/g
	Resolution	0.000001 g rms
	Resonant Freq	≥ 700 Hz
	Noise	-
	Freq Range	0.07 to 300 Hz
	Method	Piezoelectric
Author	PCB Piezotronics 393B32
	Sensitivity	5 V/g
	Resolution	0.000002 g rms
	Resonant Freq	≥ 700 Hz
	Noise	-
	Freq Range	0.1 to 300 Hz
	Method	Piezoelectric
	Author	PCB Piezotronics 393C
	Sensitivity	1 V/g
	Resolution	0.0001 g rms

Resonant Freq	≥ 3.5 kHz
	Noise	-
	Freq Range	0.01 to 1200 Hz
	Method	Piezoelectric
Author	PCB Piezotronics 626A04
	Sensitivity	10 V/g
	Resolution	0.5 μ g
	Resonant Freq	1000 Hz
	Noise	-
	Freq Range	.07 to 300 Hz
	Method	Piezoelectric

Author	Vibra.Metrics Model 1030
	Sensitivity	5 V/g
	Resolution	-
	Resonant Freq	600 Hz
	Noise	-
	Freq Range	0.07 to 200 Hz
	Method	-
Author	Wilcoxon Research 731A
	Sensitivity	10 V/g
	Resolution	-
	Resonant Freq	-
	Noise	-
	Freq Range	0.1 to 300 Hz
	Method	Piezoelectric
Author	Wilcoxon Research 731-207
	Sensitivity	10 V/g
	Resolution	-
	Resonant Freq	-
	Noise	-
	Freq Range	0.5 to 850 Hz
	Method	Piezoelectric
	Author	Hewlett Packard Company
	Sensitivity	25 V/g

Resolution	-
Resonant Freq	400 Hz
Noise	-
Freq Range	0 to 200 Hz
Method	Capacitive

Author	Fernando Garcia
	Sensitivity	0.001 V/g
	Resolution	-
	Resonant Freq	270 Hz
	Noise	-
	Freq Range	4 to 120 Hz
	Method	Piezoelectric
Author	Alessandro Bertolini
	Sensitivity	-
	Resolution	-
	Resonant Freq	0.5 Hz
	Noise	-
	Freq Range	0.01 to 150 Hz
	Method	Capacitive
Author	Felix A. Levinzon
	Sensitivity	10 V/g
	Resolution	-
	Resonant Freq	1220 Hz
	Noise	-
	Freq Range	0.04 to 800 Hz
	Method	Piezoelectric
	Author	Felix A. Levinzon
	Sensitivity	10 V/g
	Resolution	-

	Resonant Freq	370 Hz
	Noise	-
	Freq Range	0.003 to 200 Hz
	Method	Piezoelectric

Author	F. Acernese
	Sensitivity	20 V/g
	Resolution	-
	Resonant Freq	370 Hz
	Noise	-
	Freq Range	0 to 200 Hz
	Method	Interferometry
Author	Dongshan Jiang, Wentao Zhang, and Fang Li
	Sensitivity	40 dB
	Resolution	-
	Resonant Freq	630 Hz
	Noise	-
	Freq Range	5 to 400 Hz
	Method	Optical Fibber

APPENDIX B

Table 22: Optical Detection Methods Comparisons

Author	Abbaspour-Sani, E., R.-S. Huang [55]
	Sensitivity	40 mV/G
	Resolution	-
	Resonant Freq	3.2kHz
	Noise	40 mV
	Method	Optical detection using gratings
Author	Keeler, B. E. N., D. W. Carr [22]
	Sensitivity	-
	Resolution	160 fm/ $\sqrt{\text{Hz}}$
	Resonant Freq	6 to 80 MHz range depending on beam width
	Noise	-
	Method	Optical detection using gratings
Author	Olsson, R.H. [56]
	Sensitivity	-
	Resolution	50 fm/ $\sqrt{\text{Hz}}$ @ 10mHz
	Resonant Freq	-
	Noise	-
	Method	Optical detection using gratings
Author	Krishnamoorthy, U. [53]
	Sensitivity	5900V/G or 20 μm /G
	Resolution	12fm/ $\sqrt{\text{Hz}}$
	Resonant Freq	1131Hz for gimbal
	Noise	8 nG/ $\sqrt{\text{Hz}}$ (thermal noise)
	Method	Optical detection using nanogratings

<table border="1"> <tr> <td>Author</td><td>Hortschitz, W., F. Kohl [57] [112] [58]</td></tr> <tr> <td>Sensitivity</td><td>60nA/nm</td></tr> <tr> <td>Resolution</td><td>8pm/$\sqrt{\text{Hz}}$</td></tr> <tr> <td>Resonant Freq</td><td>1700Hz</td></tr> <tr> <td>Noise</td><td>80 nV/$\sqrt{\text{Hz}}$ (switched off)</td></tr> <tr> <td>Method</td><td>Optical detection using gratings</td></tr> </table>	Author	Hortschitz, W., F. Kohl [57] [112] [58]	Sensitivity	60nA/nm	Resolution	8pm/ $\sqrt{\text{Hz}}$	Resonant Freq	1700Hz	Noise	80 nV/ $\sqrt{\text{Hz}}$ (switched off)	Method	Optical detection using gratings
Author	Hortschitz, W., F. Kohl [57] [112] [58]												
Sensitivity	60nA/nm												
Resolution	8pm/ $\sqrt{\text{Hz}}$												
Resonant Freq	1700Hz												
Noise	80 nV/ $\sqrt{\text{Hz}}$ (switched off)												
Method	Optical detection using gratings
<table border="1"> <tr> <td>Author</td><td>Yao Ge, Kevin S. Kuang, Ser Tong Quek [113]</td></tr> <tr> <td>Sensitivity</td><td>300 mV/g up to 60 Hz</td></tr> <tr> <td>Resolution</td><td>-</td></tr> <tr> <td>Resonant Freq</td><td>135 Hz</td></tr> <tr> <td>Noise</td><td>-</td></tr> <tr> <td>Method</td><td>Optical Fiber</td></tr> </table>	Author	Yao Ge, Kevin S. Kuang, Ser Tong Quek [113]	Sensitivity	300 mV/g up to 60 Hz	Resolution	-	Resonant Freq	135 Hz	Noise	-	Method	Optical Fiber
Author	Yao Ge, Kevin S. Kuang, Ser Tong Quek [113]												
Sensitivity	300 mV/g up to 60 Hz												
Resolution	-												
Resonant Freq	135 Hz												
Noise	-												
Method	Optical Fiber
<table border="1"> <tr> <td>Author</td><td>Guan, T., F. Ceysens [114]</td></tr> <tr> <td>Sensitivity</td><td>1.5 mV/g</td></tr> <tr> <td>Resolution</td><td>-</td></tr> <tr> <td>Resonant Freq</td><td>1000 Hz</td></tr> <tr> <td>Noise</td><td>-</td></tr> <tr> <td>Method</td><td>Optical Fiber</td></tr> </table>	Author	Guan, T., F. Ceysens [114]	Sensitivity	1.5 mV/g	Resolution	-	Resonant Freq	1000 Hz	Noise	-	Method	Optical Fiber
Author	Guan, T., F. Ceysens [114]												
Sensitivity	1.5 mV/g												
Resolution	-												
Resonant Freq	1000 Hz												
Noise	-												
Method	Optical Fiber
<table border="1"> <tr> <td>Author</td><td>Llobera, A. [54]</td></tr> <tr> <td>Sensitivity</td><td>34.3 $\mu\text{m/g}$</td></tr> <tr> <td>Resolution</td><td>-</td></tr> <tr> <td>Resonant Freq</td><td>85 Hz</td></tr> <tr> <td>Noise</td><td>-</td></tr> <tr> <td>Method</td><td>Optical Fiber</td></tr> </table>	Author	Llobera, A. [54]	Sensitivity	34.3 $\mu\text{m/g}$	Resolution	-	Resonant Freq	85 Hz	Noise	-	Method	Optical Fiber
Author	Llobera, A. [54]												
Sensitivity	34.3 $\mu\text{m/g}$												
Resolution	-												
Resonant Freq	85 Hz												
Noise	-												
Method	Optical Fiber
<table border="1"> <tr> <td>Author</td><td>Doyle, C. and G. F. Fernando [115]</td></tr> <tr> <td>Sensitivity</td><td>1.1 $\mu\text{m/G}$</td></tr> <tr> <td>Resolution</td><td>-</td></tr> <tr> <td>Resonant Freq</td><td>592 Hz</td></tr> <tr> <td>Noise</td><td>-</td></tr> <tr> <td>Method</td><td>Optical Fiber</td></tr> </table>	Author	Doyle, C. and G. F. Fernando [115]	Sensitivity	1.1 $\mu\text{m/G}$	Resolution	-	Resonant Freq	592 Hz	Noise	-	Method	Optical Fiber
Author	Doyle, C. and G. F. Fernando [115]												
Sensitivity	1.1 $\mu\text{m/G}$												
Resolution	-												
Resonant Freq	592 Hz												
Noise	-												
Method	Optical Fiber

Author	Conforti, G., M. Brenci [116]
	Sensitivity	10 mV/μm
	Resolution	0.1 μm at 100 Hz
	Resonant Freq	100 Hz working freq
	Noise	-
	Method	Optical Fiber

Author	Zuwei, Z., W. Zhiyu[117]
	Sensitivity	130 kHz/G
	Resolution	-
	Resonant Freq	39.1 MHz
	Noise	-
	Method	Optical Fiber

Author	Schröpfer, G. [118]
	Sensitivity	1.8 V/g or 143 nm/g
	Resolution	-
	Resonant Freq	1176 Hz
	Noise	-
	Method	Optical Fiber

Author	Zandi, K. [119]
	Sensitivity	2.5nm/g
	Resolution	400 μg
	Resonant Freq	1670Hz
	Noise	-
	Method	Optical Fiber

Author	da Costa Antunes, P. F., H. F. T. Lima [120]
	Sensitivity	11 nm/G
	Resolution	-
	Resonant Freq	38.836 Hz
	Noise	-
	Method	Optical Fiber

Author Borinski, J. W., C. D. Boyd [121]	
	Sensitivity	40.8 nm/G
	Resolution	0.025 g minimum detectable level
	Resonant Freq	2000 Hz
	Noise	-
	Method	Optical Fiber
Author Wu, Y., X. Zeng [122]	
	Sensitivity	654.7 mV/G
	Resolution	-
	Resonant Freq	146 Hz
	Noise	-
	Method	Optical fiber Loop
Author Su, W., J. A. Gilbert[123]	
	Sensitivity	2 V/g
	Resolution	2.5 mg
	Resonant Freq	3500Hz
	Noise	5mV
	Method	Photoelastic effect
Author Tang, D., D. Zhao [124]	
	Sensitivity	29.4 mV/G
	Resolution	-
	Resonant Freq	4000 Hz
	Noise	-
	Method	Photoelastic Effect
Author Loh, N. C., M. A. Schmidt [125]	
	Sensitivity	39 $\mu\text{m}/\text{G}$ or 28 V/mG
	Resolution	3 mÅ/ $\sqrt{\text{Hz}}$ at 10 Hz
	Resonant Freq	80 Hz
	Noise	-
	Method	Interferometry

Author	Cooper, E. B., E. R. Post[126]
	Sensitivity	-
	Resolution	$2 \text{ mg}/\sqrt{\text{Hz}}$
	Resonant Freq	906 Hz
	Noise	$90 \text{ ng}/\sqrt{\text{Hz}}$
	Method	Interferometry
Author	Andò, B., S. Baglio [127] [128]
	Sensitivity	$6.76 \text{ \%}/(\text{m/s})$
	Resolution	-
	Resonant Freq	-
	Noise	-
	Method	Photonic Band Gap Detection

APPENDIX C

SIMULATION FOR ANGULAR MISALIGNMENT

Figure 78. Example of MEMS and CMOS misalignment

Simulations were carried out to determine the maximum angular misalignment allowed.

Given the above variable definitions (See Figure 78), a series of equations were used to simulate the error percentage produced for each angle (See below).

When $\theta=0$,

$$A_L = (X_o - \Delta x)L \quad A_R = (X_o + \Delta x)L \quad \frac{\Delta x}{X_o} \leq 0.1$$

$$L_L = X_o - \Delta x \quad L_R = X_o + \Delta x \quad L = 2\text{mm}$$

Differential signal,

$$I_L = k_1 A_L + k_2 L_L \quad I_R = k_1 A_R + k_2 L_R \quad I_{\text{Differential}0} = I_R - I_L$$

$$= 2(k_1 L + k_2) \Delta x$$

When $\theta \neq 0$,

$$A_L = (X_o - \Delta x)L / \cos\theta \quad A_R = (X_o + \Delta x)L / \cos\theta$$

$$L_L = (X_o - \Delta x)/\cos\theta$$

$$L_R = (X_o + \Delta x)/\cos\theta$$

Differential signal

$$I_L = k_1 A_L + k_2 L_L$$

$$I_R = k_1 A_R + k_2 L_R$$

$$I_{\text{Differential}\theta} = I_R - I_L = 2(k_1 L$$

$$+ k_2)\Delta x/\cos\theta$$

$$\text{Error, } \Delta I = I_{\text{Differential}\theta} - I_{\text{Differential}0} = 2(k_1 L + k_2)\Delta x(1/\cos\theta - 1)$$

From the simulation results, a series of graphs were plotted to show the relationship of the area of exposure and edge length over a range of angular misalignments of $\pm 10^\circ$. Figure 79 shows the overall results of the simulation. In order to have a 0.1% chance of error (0.001 on the graph), the maximum misalignment angle allowed would be $\pm 2.5^\circ$.

Figure 79. Overall results of simulation

APPENDIX D

LED DATASHEET

Engwinklige LED im MIDLED-Gehäuse (880 nm)
Narrow beam LED in MIDLED package (880 nm)
Lead (Pb) Free Product - RoHS Compliant

SFH 4680
SFH 4685

SFH 4680

SFH 4685

Wesentliche Merkmale

- AlGaAs-LED
- Enger Abstrahlwinkel ($\pm 20^\circ$)
- Geringe Bauhöhe
- Als Toplooker und Sidelooker einsetzbar
- SFH 4680: Gurtung als Toplooker
- SFH 4685: Gurtung als Sidelooker

Features

- AlGaAs-LED
- Narrow halfangle ($\pm 20^\circ$)
- Low profile component
- Usable as top-looking and side-looking device
- SFH 4680: Taping as Toplooker
- SFH 4685: Taping as Sidelooker

Anwendungen

- Sensorik in der Automobiltechnik
- Datenübertragung
- Fernsteuerrung

Applications

- Automotive sensors
- Data transmission
- Remote controls

Sicherheitshinweise

Je nach Betriebsart emittieren diese Bauteile hochkonzentrierte, nicht sichtbare Infrarot-Strahlung, die gefährlich für das menschliche Auge sein kann. Produkte, die diese Bauteile enthalten, müssen gemäß den Sicherheitsrichtlinien der IEC-Normen 60825-1 und 62471 behandelt werden.

Safety Advices

Depending on the mode of operation, these devices emit highly concentrated non visible infrared light which can be hazardous to the human eye. Products which incorporate these devices have to follow the safety precautions given in IEC 60825-1 and IEC 62471.

Type Type	Bestellnummer Ordering Code	Strahlstärkegruppierung ¹⁾ ($I_F = 100 \text{ mA}$, $t_p = 20 \text{ ms}$) Radiant Intensity Grouping ¹⁾ $I_e \text{ (mW/sr)}$
SFH 4680	Q65110A1570	$\geq 10 \text{ (typ. 20)}$
SFH 4685	Q65110A1571	$\geq 10 \text{ (typ. 20)}$

¹⁾ gemessen bei einem Raumwinkel $\Omega = 0.01 \text{ sr}$ / measured at a solid angle of $\Omega = 0.01 \text{ sr}$

Grenzwerte
Maximum Ratings

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Betriebs- und Lagertemperatur Operating and storage temperature range	T_{op}, T_{stg}	- 40 ... + 100	°C
Sperrspannung Reverse voltage	V_R	5	V
Vorwärtsgleichstrom, $T_A \leq 60\text{ °C}$ Forward current	I_F	100	mA
Stoßstrom, $t_p = 300\text{ }\mu\text{s}$, $D = 0$, $T_A \leq 60\text{ °C}$ Surge current	I_{FSM}	1	A
Verlustleistung $T_A = 25\text{ °C}$ Power dissipation	P_{tot}	180	mW
Wärmewiderstand Sperrschicht - Umgebung bei Montage auf FR4 Platine, Padgröße je 16 mm ² Thermal resistance junction - ambient mounted on PC-board (FR4), padsize 16 mm ² each	R_{thJA}	340	K/W
Wärmewiderstand Sperrschicht - Lötstelle bei Montage auf Metall-Block Thermal resistance junction - soldering point, mounted on metal block	R_{thJS}	180	K/W

Kennwerte ($T_A = 25\text{ °C}$)
Characteristics

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Wellenlänge der Strahlung Wavelength at peak emission $I_F = 100\text{ mA}$	λ_{peak}	880	nm
Spektrale Bandbreite bei 50% von I_{max} Spectral bandwidth at 50% of I_{max} $I_F = 100\text{ mA}$	$\Delta\lambda$	80	nm
Abstrahlwinkel Half angle	φ	± 20	Grad deg.
Aktive Chipfläche Active chip area	A	0.09	mm ²
Abmessungen der aktiven Chipfläche Dimension of the active chip area	$L \times B$ $L \times W$	0.3×0.3	mm ²

Kennwerte ($T_A = 25\text{ °C}$)

Characteristics (cont'd)

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Schaltzeiten, I_e von 10% auf 90% und von 90% auf 10%, bei $I_F = 100\text{ mA}$, $R_L = 50\text{ }\Omega$ Switching times, I_e from 10% to 90% and from 90% to 10%, $I_F = 100\text{ mA}$, $R_L = 50\text{ }\Omega$	t_r, t_f	0.5	μs
Durchlassspannung Forward voltage $I_F = 100\text{ mA}$, $t_p = 20\text{ ms}$ $I_F = 1\text{ A}$, $t_p = 100\text{ }\mu\text{s}$	V_F V_F	1.5 (< 1.8) 3.0 (< 3.8)	V V
Sperrstrom Reverse current $V_R = 5\text{ V}$	I_R	not designed for reverse operation	μA
Gesamtstrahlungsfluss Total radiant flux $I_F = 100\text{ mA}$, $t_p = 20\text{ ms}$	$\Phi_{e\text{ typ}}$	23	mW
Temperaturkoeffizient von I_e bzw. Φ_e , $I_F = 100\text{ mA}$ Temperature coefficient of I_e or Φ_e , $I_F = 100\text{ mA}$	TC_I	- 0.5	%/K
Temperaturkoeffizient von V_F , $I_F = 100\text{ mA}$ Temperature coefficient of V_F , $I_F = 100\text{ mA}$	TC_V	- 2	mV/K
Temperaturkoeffizient von λ , $I_F = 100\text{ mA}$ Temperature coefficient of λ , $I_F = 100\text{ mA}$	TC_λ	+ 0.25	nm/K

Strahlstärke I_e in Achsrichtung¹⁾gemessen bei einem Raumwinkel $\Omega = 0.01$ sr**Radiant Intensity I_e in Axial Direction**at a solid angle of $\Omega = 0.01$ sr

Bezeichnung Parameter	Symbol	Werte Values			Einheit Unit
		-R	-S	-T	
Strahlstärke Radiant intensity $I_F = 100$ mA, $t_p = 20$ ms	$I_{e \min}$ $I_{e \max}$	10 20	16 32	25 50	mW/sr mW/sr
Strahlstärke Radiant intensity $I_F = 1$ A, $t_p = 100$ μ s	$I_{e \text{ typ}}$	130	160	240	mW/sr

¹⁾ Nur eine Gruppe in einer Verpackungseinheit (Streuung kleiner 2:1) /
Only one group in one packing unit (variation lower 2:1)

Abstrahlcharakteristik**Radiation Characteristics $I_{\text{rel}} = f(\varphi)$**

Relative Spectral Emission

$$I_{\text{rel}} = f(\lambda)$$

$$\text{Radiant Intensity } \frac{I_e}{I_e 100 \text{ mA}} = f(I_F)$$

Single pulse, $t_p = 20 \mu\text{s}$

Max. Permissible Forward Current

$$I_F = f(T_A), R_{\text{thJA}} = 340 \text{ K/W}$$

Forward Current $I_F = f(V_F)$

Single pulse, $t_p = 20 \mu\text{s}$

Permissible Pulse Handling

Capability $I_F = f(\tau)$, $T_A \leq 60^\circ\text{C}$,
duty cycle $D = \text{parameter}$

Maßzeichnung Package Outlines

Maße in mm / Dimensions in mm.

Gehäuse / Package	MID mit klarem Silikonverguss / MID casted with clear Silicone
Anschlussbelegung Pin configuration	Pad 1 = Anode / anode Pad 2 = Kathode / cathode

Gurtung / Polarität und Lage

Verpackungseinheit 2000/Rolle, ø180 mm
oder 9000/Rolle, ø330 mm

Method of Taping / Polarity and Orientation

Packing unit 2000/reel, ø180 mm
or 9000/reel, ø330 mm

Maße in mm (inch) / Dimensions in mm (inch).

Empfohlenes Lötpaddesign
Recommended Solder Pad Design

SFH 4680

Padgeometrie für verbesserte Wärmeableitung
 Pad design for improved heat dissipation

Cu-Fläche > 16 mm²
 Cu-area

Lötstopplack
 Solder resist

OHF02422

SFH 4685

Padgeometrie für verbesserte Wärmeableitung
 Pad design for improved heat dissipation

Cu-Fläche > 16 mm²
 Cu-area

Lötstopplack
 Solder resist

OHF02421

Maße in mm / Dimensions in mm.

Verarbeitungshinweis: Das Gehäuse ist mit Silikon vergossen. Mechanischer Stress auf der Bauteiloberfläche sollte so gering wie möglich gehalten werden.

Handling indication: The package is casted with silicone. Mechanical stress at the surface of the unit should be as low as possible.

Lötbedingungen**Soldering Conditions****Reflow Lötprofil für bleifreies Löt****Reflow Soldering Profile for lead free soldering**

Vorbehandlung nach JEDEC Level 2

Preconditioning acc. to JEDEC Level 2

(nach J-STD-020C)

(acc. to J-STD-020C)

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Packing

Please use the recycling operators known to you. We can also help you – get in touch with your nearest sales office. By agreement we will take packing material back, if it is sorted. You must bear the costs of transport. For packing material that is returned to us unsorted or which we are not obliged to accept, we shall have to invoice you for any costs incurred.

Components used in life-support devices or systems must be expressly authorized for such purpose! Critical components¹, may only be used in life-support devices or systems² with the express written approval of OSRAM OS.

¹ A critical component is a component used in a life-support device or system whose failure can reasonably be expected to cause the failure of that life-support device or system, or to affect its safety or effectiveness of that device or system.

² Life support devices or systems are intended (a) to be implanted in the human body, or (b) to support and/or maintain and sustain human life. If they fail, it is reasonable to assume that the health of the user may be endangered.

APPENDIX E

PHOTOTRANSISTOR DATASHEET

Phototransistor im MIDLED-Gehäuse

Phototransistor in MIDLED package

Lead (Pb) Free Product - RoHS Compliant

SFH 3600

SFH 3605

SFH 3600

SFH 3605

Wesentliche Merkmale

- Enger Empfangswinkel ($\pm 20^\circ$)
- Geringe Bauhöhe (1,6 mm)
- Als Toplooker und Sidelooker einsetzbar
SFH 3600: Gurtung als Toplooker
SFH 3605: Gurtung als Sidelooker
- Emitter im gleichen Gehäuse (SFH 46xx) verfügbar

Anwendungen

- Lichtschranken, Lichtvorhänge
- Sensorik (Consumer, Industrieelektronik)
- Automobilanwendungen
- Näherungssensor

Features

- Narrow angle ($\pm 20^\circ$)
- Low profile component (1,6 mm)
- Usable as top-looking and side-looking device
SFH 3600: Taping as Toplooker
SFH 3605: Taping as Sidelooker
- Emitter in same package (SFH 46xx) available

Applications

- Interrupters, Lightcurtains
- Sensors (Consumer, Industrial applications)
- Automotive applications
- proximity sensor

Typ Type	Bestellnummer Ordering Code	Fotostrom , ($E_e=0.1 \text{ mW/cm}^2, \lambda=950 \text{ nm } V_{CE} = 5 \text{ V}$) Photocurrent $I_{pce} (\mu A)$
SFH 3600 ¹⁾	Q65110A1573	100-500
SFH 3600 -2/3 ¹⁾	Q65110A2665	100-320
SFH 3600 -3/4 ¹⁾	Q65110A2666	160-500
SFH 3605 ¹⁾	Q65110A1574	100-500
SFH 3605 -2/3 ¹⁾	Q65110A2663	100-320
SFH 3605 -3/4 ¹⁾	Q65110A2664	160-500

¹⁾ nur eine Gruppe in einer Verpackungseinheit (Streuung kleiner 2:1) / only one group in one packing unit (variation lower 2:1)

Grenzwerte
Maximum Ratings

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Betriebs- und Lagertemperatur Operating and storage temperature range	$T_{op}; T_{stg}$	- 40 ... + 100	°C
Kollektor-Emitterspannung Collector-emitter voltage	V_{CE}	35	V
Kollektorstrom Collector current	I_C	15	mA
Kollektorspitzenstrom, $\tau < 10 \mu s$ Collector surge current	I_{CS}	75	mA
Emitter-Kollektorspannung Emitter-collector voltage	V_{EC}	7	V
Verlustleistung, $T_A = 25 \text{ °C}$ Total power dissipation	P_{tot}	130	mW
Wärmewiderstand Sperrschicht/Umgebung ¹⁾ Thermal resistance junction	R_{thJA}	340	K/W
Wärmewiderstand Sperrschicht/Lötstelle ²⁾ Thermal resistance junction	R_{thJS}	180	K/W

¹⁾ bei Montage auf FR4 Platine, Padgröße je 16 mm² / when mounted on PC-board (FR4), pads size 16 mm² each

²⁾ bei Montage auf Metall-Block / when mounted on metal block

Kennwerte ($T_A = 25 \text{ °C}$)
Characteristics

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Wellenlänge der max. Fotoempfindlichkeit Wavelength of max. sensitivity	$\lambda_{S \max}$	990	nm
Spektraler Bereich der Fotoempfindlichkeit $S = 10\%$ von $S_{\max}$ Spectral range of sensitivity $S = 10\%$ of $S_{\max}$	λ	500 ... 1100	nm
Bestrahlungsempfindliche Fläche Radiant sensitive area	A	0.04	mm ²
Abmessungen des Chips Dimensions of chip	$L \times B \times H$ $L \times W \times H$	0.35×0.35 $\times 0.185$	mm × mm × mm

Kennwerte ($T_A = 25\text{ °C}$)

Characteristics (cont'd)

Bezeichnung Parameter	Symbol Symbol	Wert Value	Einheit Unit
Halbwinkel Half angle	φ	± 20	Grad deg.
Kapazität, $V_{CE} = 5\text{ V}$, $f = 1\text{ MHz}$, $E = 0$ Capacitance	C_{CE}	1.3	pF
Dunkelstrom Dark current $V_{CE} = 20\text{ V}$, $E = 0$	I_{CEO}	1 (≤ 50)	nA

Die Fototransistoren werden nach ihrer Fotoempfindlichkeit gruppiert und mit arabischen Ziffern gekennzeichnet.

The phototransistors are grouped according to their spectral sensitivity and distinguished by arabian figures.

Bezeichnung Parameter	Symbol Symbol	Wert Value			Einheit Unit
		-2	-3	-4	
Fotostrom, $\lambda = 950\text{ nm}$ Photocurrent $E_e = 0.1\text{ mW/cm}^2$, $V_{CE} = 5\text{ V}$	I_{PCE}	100 .. 200	160 ... 320	250 ... 500	μA
Anstiegszeit/Abfallzeit, $\lambda = 950\text{ nm}$ Rise and fall time $I_C = 100\text{ }\mu\text{A}$, $V_{CC} = 5\text{ V}$, $R_L = 10\text{ k}\Omega$	t_r , t_f	30	45	70	μs
Kollektor-Emitter-Sättigungsspannung Collector-emitter saturation voltage $I_C = I_{PCEmin}^{1)} \times 0.3$, $E_e = 0.1\text{ mW/cm}^2$, $\lambda = 950\text{ nm}$	V_{CEsat}	150	150	150	mV

¹⁾ I_{PCEmin} ist der minimale Fotostrom der jeweiligen Gruppe / I_{PCEmin} is the min. photocurrent of the specified group.

Directional Characteristics

$$S_{\text{rel}} = f(\varphi)$$

Rel. Spectral Sensitivity, $S_{\text{rel}} = f(\lambda)$, axial direction

Dark Current $I_{\text{CEO}} = f(V_{\text{CE}}), E = 0$

Dark Current $I_{\text{CEO}} = f(T_A), V_{\text{CE}} = 20 \text{ V}, E = 0$

Photocurrent $I_{\text{PCE}} = f(T_A), V_{\text{CE}} = 5 \text{ V}$, normalized to 25 °C

Collector-Emitter Capacitance $C_{\text{CE}} = f(V_{\text{CE}}), f = 1 \text{ MHz}$

Maßzeichnung Package Outlines

Maße in mm (inch) / Dimensions in mm (inch).

Gehäuse package	MID mit klarem Silikonverguss MID casted with clear Silicone
Anschlussbelegung Pin configuration	Pad 1 = Kollektor / collector Pad 2 = Emitter / emitter

¹⁾ Verarbeitungshinweis:

Das Gehäuse ist mit Silikon vergossen. Mechanischer Streß auf der Bauteiloberfläche sollte so gering wie möglich gehalten werden.

¹⁾ Handling indication:

The package is casted with silicone. Mechanical stress at the surface of the unit should be as low as possible.

Empfohlenes Lötpaddesign Recommended Solder Pad Design

Verarbeitet als Toplooker
Mounted as toplooker

SFH 3600

Padgeometrie für
verbesserte Wärmeableitung

Pad design for improved
heat dissipation

Cu-Fläche > 16 mm²
Cu-area

Lötstopplack
Solder resist

OHF02422

Verarbeitet als Sidelooker
Mounted as sidelooker

SFH 3605

Padgeometrie für
verbesserte Wärmeableitung

Pad design for improved
heat dissipation

Cu-Fläche > 16 mm²
Cu-area

Lötstopplack
Solder resist

OHF02421

Lötbedingungen**Soldering Conditions****Reflow Lötprofil für bleifreies Löten****Reflow Soldering Profile for lead free soldering**

Vorbehandlung nach JEDEC Level 2

Preconditioning acc. to JEDEC Level 2

(nach J-STD-020C)

(acc. to J-STD-020C)

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APPENDIX F

Figure 80. Electrical diagram for device

Figure 81. Breadboard prototype with zoom in of location to place MEMS device

The basis for the electrical layout is based on the same design used by Hortschitz [57, 58, 112], which has enabled him to produce the results he has achieved (See Figure 80). For this, a breadboard prototype for the electronic circuitry was developed and has been tested to working efficiently without the use of the MEMS layer (See Figure 81). The datasheet for both intensity photodetector and LED are given in APPENDIX D and APPENDIX E. The entire breadboard design would be converted into a printed circuit board (PCB) prototype and subsequently the PCB with the MEMS device would be used for testing of human footsteps. The fabricated device would be placed in between the photodetector and LED portion as given in Figure 82, not drawn to scale, with a holder placed in between to secure the MEMS in place.

Figure 82. PCB diagram for final device