BME 4910 FINAL REPORT

THERMOACOUSTIC CELL SORTER

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April 22, 2010

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ABSTRACT

There exists a need for an in-vivo cell sorting device for the application in the medical industry. This could be used on site to remove harmful or perhaps cancerous cells. Current methods used for cell sorting require bulky and costly machinery. Proposed here is a Thermoacoustic based cell sorter to validate certain novel components to be eventually used in an in-vivo device.

The primary novel subunits that constitute this device include a cell recognition technique based on the thermoacoustic principal and an actuating micro-fluidic valve composed of a hydro-gel matrix composite with dispersed alumino-silicate zeolite particles. Proof of concept requires the preferential sorting of cell simulating colored particles in a singular channel flow system.

To accomplish this various alternative designs will be considered and an optimal design will be selected. Modeling will be performed to simulate flow through the system. Primary parts will then be obtained in the most cost-effective manner. A number of prototypes of each sub-system will be constructed and tested on before the assembly of the entire system. The final assembly will include the integration of all sub-units both mechanically and electronically. A software program will be used to control read the output from the cells as well as to control the voltage actuating valve.
1 INTRODUCTION

1.1 BACKGROUND

The thermoacoustic cell sorter or, TACS, is a research based device for Dr. Shiva Kotha. Dr. Kotha is part of the Biomechanics, Rehabilitation Engineering and Ergonomics disciplines of the University of Connecticut’s Biomedical Engineering program. Completion of this device will allow for further research for Dr. Kotha and his team of graduate students. There are components to this project that will be incorporated in the device that need to be newly developed by the members of this team. These materials could also prove useful to engineering world. The product is to be cost effective and energy efficient, without posing major obstacles to biocompatibility.

1.2 PURPOSE

There are a variety of purposes for this project. The main goal for the this project will be the ability for Dr. Kotha and his team to further their research and fine tune the device for later commercial use. Although this device will be less efficient than currently used cell sorting devices, the much smaller size will allow for implantation inside a human body. The special part about this project is that the future devices based upon the TACS will have a potentially large impact on the human population. Upon successful implantation, this will allow for the filtration of specifically targeted cells. The most likely location for the implantation of this device will be the circulatory system. If implanted in the circulatory system, the device could potentially remove harmful cells and particles from the bloodstream, or be used to administer drugs to targeted cells, rather than allowing diffusion of the drugs into all cells including those which would be unaffected or damaged by the drug. The device could also be used to sequester a sample of specific cells from the blood for further investigation, without having to remove an entire blood sample then separate the cells. If implanted in the lymphatic system, the filtration and isolation of cancerous cells would be the final goal.

The device needs to be constructed in such a way that the proof of concept will enable construction of the device on a scale appropriate for in vivo use. Traditional cell sorting devices use a combination of an optically based data gathering technique, and electromagnets to physically move the cells into different bins. The electromagnetic sorting would not be applicable inside the human body. This necessitates the use of the actuating gates inside the microchannel, a considerably slower method. Because our sorting rate is now limited by the actuating gates, it allows us to use a less efficient cell identification procedures and data analysis technique based on the thermoacoustic principle.

1.3 PREVIOUS WORK DONE BY OTHERS

The TACS has many components, some of which will be incorporated into the project from previous research, and other components that are currently being researched by our team for use in the final device. The incorporation of such a variety of technologies sets the TACS apart from other cell sorters. Traditional cell sorting devices use an optically based approach to gather information about cells and identify them. These systems use the concept of fluorescence in order to sort and gather information about the cell. Fluorescence is the ability of a molecule to absorb energy, typically light energy, and shift
to an excited state. After reaching the excited state, the molecule releases the energy at specific wavelengths. Detectors placed around the cell measure the intensity of different wavelengths to determine whether or not certain molecules are present in the cell. The current process of examining and counting cells is known as flow cytometry.

### 1.3.1 PRODUCTS

Invitrogen is a Biotechnology company that sells a flow cytometry device. The device differs from the TACS because it uses a light based method to gather data on the flowing cells. In this device, the cells pass through a laser beam which excites the cells. During this process the device simultaneously uses an optical detector to capture the light which passes through and is scattered by each cell. The light captured enables the system to gather information about the size, structure, phenotype and integrity of the cell. With the information gathered, the flow cytometer can help in the fields of immunophenotyping, ploidy analysis, cell counting, and GFP expression analysis. This technique has the capability to sort thousands of cells per second due to the high flow rate.

The intensity of the detected light and type of light scattering caused by each cell is determined by its characteristics. To sort the cells, fluorescent antibodies are attached to cells. Since the antibodies attach to known cells we are able to sort them. Band pass filters are used in order to gather only the desired fluorescing wavelengths that are produced by the excitation of the laser.

The device is composed of several parts. The first of which is the fluidic system, this is where the cells flow in a single file line and are exposed to the laser. Secondly is the laser itself, this system sorts the cells based on their excited properties due to the laser. Thirdly, the optics direct the scattered light and fluorescing colors to the fourth system, the detectors. The detectors send the information to the last system. The last system is the network of peripherals and the computers which digitally display and further analyze the data [1].

### 1.3.2 PATENT SEARCH RESULTS

The patent that is already in place for cell sorting is the United States Patent 7160730. The methods that are patented use the single flow cell fluidic channel with a laser as the mode of excitation. Once excited and fluoresced the different scattered light rays from the cells are detected and analyzed. The specific fluoresced cells are then divided into different ports of the desired color or property. Since this method uses the light rays as the mode of detection, rather than piezoelectric sensors, it is different from the TACS which uses the frequencies given off by the fluorescing cells [2].

A patent application, number 12/334,217, describes a system for differentiating between cells in vivo by detecting photoacoustic waves. The device described by this patent is sufficiently different from ours, as it is noninvasive, it is specifically designed for use in living tissues, and there is no method for separating target cells from the flow. It only includes a method for killing targeted cells [3].
1.4 MAP FOR THE REST OF THE REPORT

The TACS device contains multiple subcomponents. In the following pages, we will describe how the device functions, and thus the document is often broken up into sections pertaining to each of the three subcomponents. In the following pages, information about alternative designs as well as the optimal design of the TACS device can be found. We go on to describe the realistic constraints of the project, safety issues associated with the device, the impact we expect the device to have on the engineering community and the world, and what we have learned from being teammates on this project.

As far as the management aspect of this project is concerned, our budget and timeline is presented along with a detailed list of the contributions of the three team members and our acknowledgements.

The final sections of this document include all technical references and documentation associated with the purchased components of this product.
PROJEC\n\n\n\n1.5 OPTIONS CONSIDERED IN THE DESIGN OF THIS DEVICE

1.5.1 FLOW CONTROL THROUGH THE DEVICE

Before any information can be gathered about the cells, and before any sorting takes place, the cell stock needs to be contained in a proper fashion. To accomplish this, the unsorted cell stock will be held in a container of some sort. Originally, an IV bag was going to be used for our design. The benefits for using an IV bag involve the slow flow rate at which the fluid leaves the bag. Furthermore there is less tubing required to direct the flow from the bag to the capillary tubes when comparing to using laboratory bottles. There are also products on the market that help regular flow at specific rates when using an IV bag. There are three different approaches that can be used when considering the attachment of the saline bag to the capillary tubes. The different methods affect the cellular flow for the TACS device in different ways.

1.5.1.1 RUBBER STOPPER

The first approach to be taken is simply to use a rubber stopper device that will allow for the insertion of the capillary tube directly into a short piece of standard tubing coming from the IV bag. The rubber device will be tapered outward from the insertion location of the tubing and have an opening for the .665 millimeter diameter capillary tube. This approach will be the most cost efficient method but may create more problems in the future. This does not directly provide us with a flow rate before we start testing, nor does it provide us with a way of regulating the flow rate once the device is complete. If this rate is too slow, or too fast, we will need to purchase and incorporate a flow regulating device to manage the problem later in the design process, this situation is not ideal.

1.5.1.2 FLOW REGULATOR

Another approach allows us to regulate the flow directly by using an IV flow regulator. These devices are used in hospitals to regulate the delivery of medicines that are contained in saline bags. An IV flow regulator can directly regulate the cellular flow anywhere from 0 to 250mL per hour. Such devices use a dial to set the rate. Knowing the flow rate of the system is crucial for the success of the device. The drawback of this apparatus is that IV flow regulators potentially require a physician’s license to purchase. The cost is not yet known for one of these devices but a response is imminent.

1.5.1.3 HYDRODYNAMIC FOCUSING

The third and final approach would be to use a hydrodynamic focusing device. Such devices use a sheath or saline solution that enters and mixes around the sample solution at a faster velocity.
The faster moving liquid applies a pressure to the cellular stock in accordance with Bernoulli’s principle. The disadvantage to this device is that we would have to make one and it can not simply be purchased. All of these devices provide specific advantages and disadvantages that may help or hinder the progression of the TACS. The final destination for the sorted groups of cells will be a saline bag just like the one at the start of the system. The volumes will all be the same so one bin will be able to accommodate the complete cell solution.

Due to the clients request a different approach for the flow holder was needed. Rather than using an IV bag, a series of plastic laboratory bottles will be used to hold the cellular stock. This approach allows for the user to reuse the cellular stock holder. To achieve this, bottles will be ordered and fused together combining both opening ends. This method will require more tubing and a valve in order to stop and start the flow. The flow mechanisms are going to be used are described in greater detail in the optimal design portion of the report.

1.5.2 ELECTRONIC CELL DIFFERENTIATION METHOD

The method by which cells are identified as belonging to one group or another consists of a laser diode to excite and heat the cells, PVDF films of different thicknesses to detect the resulting ultrasound wave, and some system to interpret the voltage response from the PVDF films and then send a signal to the correct gate to open or close. In the following sections I will discuss the different options for each part.

1.5.2.1 LASER DIODE

The choice of the laser diode is largely based on the type of sorting which we would like to do. In different versions of the device, the laser diode is likely to change. In order to maximize the ability of the
system to differentiate between cell types, we need to maximize the differences between the voltage responses given by the PVDF films, that is, we need to maximize the differences between the longitudinal waves given off by each cell type. The only way to do this is by selecting a laser diode with a wavelength which will greatly excite the molecules in one cell type, but not in the other cell type.

To illustrate this point, I’d like to look at red blood cells, as they are commonly the subject of photoacoustic imaging. Consider the case of oxygenated blood and deoxygenated blood. Oxygenated blood appears red, which means that it reflects red light, whereas deoxygenated blood absorbs red light. If a red laser diode were used in this case, the molecules in deoxygenated blood would absorb more energy, whereas the molecules in oxygenated blood would reflect the majority of the incoming light. Therefore, more thermal expansion would occur in the blood cell devoid of oxygen, causing the longitudinal wave to have greater amplitude, and the resulting voltage response of the ultrasound sensors to be greater than those resulting from the oxygenated blood cell.

![Figure 2 - Absorption Spectra of Oxygenated and Deoxygenated Hemoglobin [5]](image)

As shown in Figure 2, [5], oxygenated hemoglobin absorbs much less energy from light with a wavelength between 600-700nm.

An example test for this system could be to differentiate between living and dead cells. A common method of doing so would be to expose the cells to trypan blue dye, which traverses a dead cell’s membrane but not a living cell’s. Trypan blue dye has an absorption peak at 580nm [8], which is yellow light. The closer the laser diode’s wavelength is to this absorption peak, the greater the thermal expansion of the dead cells would be. Unfortunately, laser diodes are only available at discrete wavelengths, and none of the currently available laser diodes are yellow. The closest available wavelengths of laser diodes would be 532nm (green) or 635nm (red). 650nm laser diodes are the most common laser diodes, and the least expensive.

The 650nm laser diode is the least expensive and easiest to acquire laser diode. A potential drawback of this option is that it may not cause a noticeable difference in thermal expansion between cell types.

Both the green (532nm) and red (635nm) diodes mentioned before are about 50nm off-center from the absorption peak of trypan blue. So both diodes would likely produce the same devices of the
same accuracy. These devices would be more accurate than a device which uses a 650nm diode, but would cost a bit more.

There is no reason why we could not place several diodes one after another on the capillary tube and measure the thermal expansion of the cells in response to different frequencies of light. Naturally this would be most expensive option. This would also slow down the decision making software as it would increase the amount of input which would need to be interpreted and evaluated. This would also likely be the most accurate option.

### 1.5.2.2 PVDF FILMS

The PVDF films are relatively expensive. The thinner the film is, the more expensive it is. The commonly found thicknesses for PVDF films are 110 micron, 52 micron, 28 micron, and 9 micron. 9 micron film is almost as expensive per square cm as the other three thicknesses combined. The frequency range to which each film responds is related to the thickness of the film by the following equation, $F_{\text{max}} = \frac{V}{2\pi t}$ [6]. Where $V$ is the speed of sound in the film, $t$ is the thickness and $F_{\text{max}}$ is the frequency at which the film resonates.

Our options for this component range from using a single sensor, to using one of each sensor.

Using a single sensor made of the 110 micron film would naturally be the least expensive approach, but also likely the least effective.

Using one sensor of each thickness would provide the most information about the cells, and definitely yield a better data set for differentiating between cell types, but also cost the most. This would also increase the amount of computing time required to evaluate the signals.

Once we’ve tested each PVDF sensor, it will be easier to tell which signals actually contribute to the overall accuracy of the device, and which are recording the most noise. If we find that one of the sensors provides a signal which clearly differentiates between the two cell types then we could afford to use only one sensor and save money on manufacturing, and computing time.

### 1.5.2.3 ELECTRONIC CLASSIFICATION SYSTEM

The classifier has the following requirements:

- Read the voltage signal from the PVDF films
- Determine the type of cell
- Output a voltage to open and close voltage actuated gates

Both design options will include a certain amount of circuitry in order to ramp up or ramp down the voltage responses of the PVDF films, and filter out environmental noise.
1.5.2.3.1 CIRCUITRY BASED OPTION

A circuitry-based design of the decision making module would likely cut the amount of money spent on parts, but it would not allow the use of complex signal analysis. This method would also complicate debugging, as the number of components could become very large.

1.5.2.3.2 SOFTWARE BASED OPTION

A software-based approach to solving our problem would be more expensive; it would necessitate the purchase of a data acquisition device, as well as compiling a LabVIEW or Matlab program. In our case, these are already at our disposal, but in industry, this would increase the cost of materials. This approach allows us to use complicated functions such as Fourier transforms, wavelet transforms, etc. which could provide huge increases in the accuracy of the device. Debugging the programs will be considerably simpler than debugging a complicated circuit board. So this method would be less time consuming, and yield a more accurate device.

1.5.3 GATE MECHANISM

There exists a need to separate the particles into the appropriate bins. A gate will need to be designed to regulate the flow into appropriate bins. Since the overall apparatus is small, a gate that can switch between open and closed with electrical stimulation would be ideal.

1.5.3.1 ACTUATING ZEOLITE MATERIAL

The first design for the gate involves the use of a nano-porous Zeolite material. This material will be impregnated with copper. Upon an applied electric field, the material will absorb ions into the nano-pores. These ions will come from a hydro-gel that will then contract and decrease in volume. This contraction will cause the material to un-block the flow to a certain collection bin, thus adequately separating the particles.

![Figure 3 - Charged Nanopore of Zeolite Crystals](image)

**Figure 3 - Charged Nanopore of Zeolite Crystals [7]**
There are certain drawbacks of this design. This material has not been used as a gate material previously and the performance in such an application is unknown. In addition, the preparation methods for the zeolite material could prove costly and troublesome. The advantage are that this is a low cost and low energy design. Additionally, hydrogels are known to be very biocompatible which could help with our final goal of implantation in vivo.

1.5.3.2 MAGNETIC GATE

An alternative design for a flow control mechanism involves the use of a magnetic material. At the correct moment, an electric field will be applied and the material will acquire a magnetic moment. It will be directed with an opposing permanent magnet to a particular edge of the capillary tubing. It will thus block the flow in one direction. This method could either stop the flow in one capillary or direct the flow between two capillaries. This then allows for particle sorting. The drawback of this method is that securing the magnetic material in place may be challenging.

1.5.3.3 THERMALLY EXPANDING GATE

A third method is to use a shape memory material such as nitinol as the gate or a thermally expanding material. An electric current would be applied to supply heat and the material would expand to block the flow. It would then return to its original shape upon cooling. The drawback of this method is that it may take time for the material to heat and cool. This does not allow for fast actuation of the gate and thus limits the amount of particles that can be sorted per unit time.

1.5.3.4 COMMERICALLY AVAILABLE GATE

A final method is to use a commercially available gate. A search for commercially available products was conducted. An ideal solution was found to use the Dolomite Company based in Europe. The gate actuates in a very timely fashion (8ms), and come with simple connections and instructions for use. This valve employs a relatively low voltage (12V) which leads to improved safety of the product. Additionally, it is an in stock item and can be shipped out of the Massachusetts branch in 24 hours.

The size allows valve mounting areas and flow path lengths to be minimized. The PEEK body and perfluoroelastomer diaphragm ensure that fluid only meets with inert materials. This could offer an energy efficient, although perhaps a costlier solution.

Figure 4 - Commercially Available Microfluidic Valve [12]
1.5.3.5 ELECTRO-WETTING GATE

A method for a micro-fluidic gate that has already been researched and implemented is a gate based on the principal of electro-wetting. Some materials are hydrophobic and can be induced to changing to hydrophilic upon the application of a voltage.

![Diagram of an Electro-wetting Microfluidic Gate](image)

Figure 5 - Diagram of an Electro-wetting Microfluidic Gate [11]

Before the voltage is applied, the fluid does not flow due to the hydrophobicity of the material. Once a voltage is applied, the water based flow can pass. The advantage of this design is its ease of use and fast actuation time. The primary disadvantage of this design includes expensive or difficult fabrication.

1.6 OPTIMAL DESIGN

1.6.1 OBJECTIVE

The project presented here is a device which can sort cells based on their colors. It has been requested by our client Dr. Shiva Kotha at the University of Connecticut. Although there are currently cell sorting devices for sale on the market, the device designed here will accomplish cell sorting by applying the thermoacoustic principle. The thermoacoustic principle states that molecular bonds can be excited by incoming electromagnetic radiation. The energy contained in the waves is converted to thermal energy which spreads to the surrounding particles. Heating of these particles causes thermal expansion. Expansion causes a displacement of fluid particles near the object, and thus an acoustic pressure wave is emitted. The use of light to excite the molecular bonds is called the photoacoustic principle. Depending on the color of the molecules, especially targeting dyes, in the cells and the wavelength of the light used to excite the cells, different acoustic waves will be emitted. It is from the emitted energy waves that the
The thermoacoustic cell sorter, or TACS, will be able to detect differences in the cells flowing through the device and eventually sort them.

This device uses methods that have not yet been used for cell sorting. The thermoacoustic principle is the first aspect of this device that stands out. Other devices use optic detection to determine the properties of cells. We are also using polyvinylidene fluoride (PVDF) films to detect the ultrasound signals. Secondly, the use of an actuating Zeolite gate will direct the flow of cells to the appropriate sorting bin. There are three main parts to this design. The flow control system will allow for the correct rate and proper movement through the device. Secondly, the electronics system will essentially excite the cells and detect the ultrasonic waves given off by each of the colored cells. Another task for the electronics will also be to analyze the ultrasonic data and be able to send a signal to the final part of the TACS device, the actuating material. The Zeolite actuating material will be controlled by a voltage. This will either open or close the gate based on the decision making capabilities of the electronic system. By proper incorporation of the three unique systems the device will be able to sort cells using methods not currently being used or tested in the cell flow cytometry field. The main scope of this project is to research to see if using these methods can truly work for sorting cells.

1.6.2 SUBUNITS

1.6.2.1 DEVICE FLOW CONTROL

The TACS system starts out with a cellular stock containing unsorted cells. After reviewing the three previous flow control designs it was decided not to use any of them. Instead, in place of an IV bag the TACS device will use a plastic laboratory bottle to hold the unsorted cell stock. The bottle was decided upon because this will allow for cleaning and reuse of the component. In order to make the device user friendly the bottle will have openings at both the top and the bottom. The top side will have a standard wide-mouth opening to allow the unsorted cell stock to be poured in. The bottom opening of the bottle will have two options for caps. One cap will also be a standard closing cap if the cell stock needed to be transported to a different location. The second cap will be connected TACS device itself. The cap will contain an output opening in which a hose or tube like apparatus can be attached. Standard tubing will be used to start the flow of the system. In order to enable and disable the flow a value will be incorporated into the standard tubing apparatus. The standard tubing is flexible it will allow for more movement and make it easier to connect the various parts. The standard tubing can then be inserted into the various microchannel molds. There will be two containers that are much larger than the one used for cell holding that contain saline solution. The two containers will have tubulation outlets attached two standard tubing and a nozzle. These two containers will each be attached to a straight channel magnetic gate. The two saline containers will act as a means to direct the flow through an H-Gate. H-Gates have the sample input and two other flow inputs which direct the center sample flow to one of two output channels depending on which flow is faster. H-Gates allow for this flow redirection based on micro fluidics and the Bernoulli’s principle, they also eliminate the risk of the gates closing on the flowing cells. Since the microchannels are fabricated using flat patterns the ultrasound will not have any problems with refraction when leaving the excited particles.
Ideally the pressure of the liquid in the holding bottles would force the fluids through the system. Before the flow starts it is required that the top cap is removed to allow air out of the bottle. This will create the pressure difference needed to force the liquid stock down through the tubing and completely through the system. Bernoulli’s principle states that as the velocity of a moving fluid increases the pressure exerted by the fluid decreases. Since the cellular stock in the system is liquid, it will follow Bernoulli’s law of incompressible flow. This law allows us to assume that the density of the liquid is a constant throughout the system regardless of pressure. Because our test particles are incompressible solids, and in the future cells are assumed to consist wholly of water, this assumption will be applicable in our calculations. Bernoulli’s equation in its original form is as follows:

\[
\frac{1}{2} \delta \nu^2 + \delta gh + p = \text{cons} \tan t
\]

In this equation \(\delta\) is the density of the fluid which is expected to be constant throughout the system, \(\nu\) is the velocity of the flow at a given point. Lowercase \(g\) is the gravitational acceleration constant (9.84/s^2 or 32.2ft/s^2), \(h\) is the height from a reference level. \(P\) is the static pressure of the liquid [13]. From the equation we see that it equals a constant value. Since the pressure due to height difference is extremely small compared to the other values we can further simplify the equation to:

\[\text{Static Pressure} + \text{Dynamic Pressure} = \text{cons} \tan t\]

Since the areas of the standard tubing to microchannel will change throughout the system, so will the velocity of the fluid flow. By the principles of mass conservation we see that:

\[\text{Flow}_A_1 = A_1 V_1 \Delta t\]
\[\text{Flow}_A_2 = A_2 V_2 \Delta t\]
\[\text{Flow}_{\text{Mass}}_1 = \delta A_1 V_1 \Delta t\]
\[\text{Flow}_{\text{Mass}}_2 = \delta A_2 V_2 \Delta t\]
\[\text{MassConservation}\]
\[A_1 V_1 = A_2 V_2\]

From the final step of the equations simplification we can see that when the area of the channel or tube decreases, the velocity of the fluid through that area increases. Knowing this allows us to design the flow system accordingly and hopefully be able to measure the appropriate flow rate. The flow rate is critical to the success of this device, if we know how fast the cells are flowing through the microchannel we can set a time delay in the LaVVIEW program in order to tell the gate to open or close at the correct time.

Once the flow from the standard tubing reaches the microchannels the velocity will change. Multiple trial runs and the Bernoulli’s equation will be used to determine the velocity through the capillary tubing. This part is critical for the system so the timing of the flow, electronics, and gate can all work accordingly. Another issue with the flow that will need to be tested will be the flow through the actuating gate. If flow can not be induced simply by the pressure the holding containers apply for force, a syringe can be used to the first stages of testing for the device. If TACS does not reach the stage in which
the tubing is all connected, syringes can be used to test the different methodologies of flow that want to be employed for this project.

### 1.6.2.2 CELL DIFFERENTIATION SYSTEM

The main goal of the project is to sort cells electronically using sensors which can be made on a microscale for implantation. In order to achieve this, we will make use of a specific form of the thermoacoustic principle, the photoacoustic principle. The thermoacoustic principle states that materials emit acoustic waves when they undergo thermal expansion and contraction. The specific use of light to cause localized heating to drive said thermal expansion is called the photoacoustic principle. The photoacoustic principle is employed in noninvasive imaging, and involves the use of a laser to enervate soft tissue and ultrasound sensors to measure the acoustic output of the tissue. Reconstruction algorithms can then generate an image based on the strength, direction, and phase of the acoustic waves [9]. In the TACS system, certain features of the photoacoustic waves will be measured and calculated as a means to differentiate between different cell types rather than constructing an image.

Cells in the TACS will be energized by a laser diode. Certain molecular bonds will absorb the energy of the laser diode and heat up. The localized heating will cause rapid thermal expansion in the cell. When the cell expands, it will displace fluid on all sides, causing an increase in pressure. Heat will then be dissipated to the surrounding fluid, and the cell will shrink to its previous size, pulling fluid molecules towards itself and thus causing a decrease in pressure. The rapid increase and decrease in pressure is otherwise known as an acoustic wave, and will likely have a frequency in the ultrasonic range [10]. The acoustic waves will be recorded by ultrasound sensors mounted directly on the outer walls of the capillary tubes. The exact configuration of these sensors will be determined by experiment in order to optimize the signal to noise ratio.

In order to make electronic classification of the cells as easy and accurate as possible, the TACS is being designed in such a way that the voltage response of the ultrasound sensors will vary greatly based on which type of cell is in under inspection. This involves choosing a laser diode which will cause varying levels of thermal expansion based on the makeup of different cells, as well as choosing ultrasound sensors which are tuned to the frequency of the propagated acoustic waves. Dyes and other indicators can be used to improve a cell’s thermal response to a certain frequency of laser light.

### 1.6.2.2.1 LASER DIODE AND DRIVER

The factors we have considered in the selection of the laser diode for the TACS include the following:

1. Wavelength of light
2. Pulse energy
3. Pulse duration

The best wavelength of light is in the red or infrared spectrum. This is because these wavelengths are best converted to heat by living cells. A wavelength in this range will also serve us well when...
inspecting the colored beads, as some will respond more to the red or IR spectrum than other. An SPL PL85 laser diode from Opto Electronics will be used to enervate the test specimens. This laser diode operates at pulse widths from 1 ns to 100 ns, which enables a range of configurations to modulate the energy of the laser pulse. The laser diode will emit a frequency of light of 850 nm. The ideal repetition rate and pulse width will be determined by experiment, and configured later.

The ETX-10A is a laser diode driver compatible with the SPL PL 90 laser diode. The laser diode driver has TTL modulation which enables us to trigger the pulse programmatically. This laser diode driver circuit also allows modulation of the output voltage and current, this will prove useful if we choose to use another laser diode and need to reconfigure the driver.

1.6.2.2.2 ULTRASOUND SENSORS

The ultrasound sensors integrated in the TACS are made from polyvinylidene fluoride (PVDF), a piezoelectric material. Piezoelectric materials are materials which have all the dipoles, i.e., charged particles, in them aligned. Because of this alignment, any motion which causes bending or pressure changes in the material will distort the density of dipole moments in the material, thereby producing a voltage.

We have chosen to construct the ultrasound sensors with PVDF because of its many beneficial properties. PVDF and its copolymers are known to be the most highly piezoelectric organic compounds; it produces a voltage output up to ten times higher than a piezoelectric ceramic given the same applied pressure. Its dipole is considerably more stable than comparable ceramics, as it can withstand electric fields of up to 75 V/μm. PVDF is also resistant to moisture damage and most forms of radiation, and it is very chemically inert. The use of a polymer also allows us to bend the material and adhere it to the capillary tube using ordinary adhesives [6]. The PVDF films have been acquired from Precision Acoustics because their pricing packages were the most cost effective.

![Figure 6 - Electrical Representation of a PVDF Film [6]](image-url)
The PVDF films can be modeled as an alternating current source in series with a capacitor as shown in the figure above.

In order to ensure accuracy in signal processing, the TACS will employ 4 different ultrasound sensors, constructed from PVDF films of different thicknesses. The PVDF films will have thicknesses of 9 micron, 28 micron, 52 micron, and 110 micron. Each sensor acts as a first-order filter with a maximum response at

\[ F_{\text{max}} = \frac{v}{2\pi t} \]  

where \( v \) is the speed of sound in the film (\( 2.2 \times 10^3 \text{ m/s} \)), and \( t \) is the thickness of the film. So the center frequencies of the four films will be approximately 40 MHz, 20 MHz, and 1 MHz. The combination of frequencies will provide the data processing unit the necessary information to differentiate between cell types.

The PVDF films will be inspected visually, and tested using an oscilloscope and high frequency vibrations produced by an 1 MHz ultrasound transducer. This frequency matches falls close enough to the center frequencies of the sensors so it will provide an adequate test.

PVDF is not only piezoelectric, it is also pyroelectric, meaning it will produce a voltage in response to a change in temperature. In future applications, this will not be a necessary concern because of homeostasis inside the body. The voltage response, \( V \), caused by a change in temperature in degrees Celsius, \( \Delta T \), is given by the equation:

\[ V = \frac{p}{T} \Delta T / \varepsilon \]

where \( p \) is the pyroelectric coefficient of the film, lowercase \( t \) is the thickness of the PVDF film in meters, and \( \varepsilon \) is the permittivity of the film. The permittivity of PVDF is \( 106 \times 10^{-12} \text{ C/Vm} \), and the pyroelectric coefficient is \( 30 \times 10^{-6} \text{ C/m}^2 \) [6]. As shown by the equation above, the pyroelectric voltage response is proportional to the change in temperature. Temperature fluctuations in the design lab are likely going to be at a very low frequency and should produce a voltage response which is very low frequency. So a high pass filter should effectively eliminate the pyroelectric effect on the voltage response.

Sensor circuits are also vulnerable to the electromagnetic fields induced by the alternating current of the domestic power supply, which oscillates 60 Hz. This is also well below the expected frequency response of the filters.

A simple high pass filter will effectively remove these sources of noise before signal processing takes place.

Analog filters will be tested by connecting a sine function generator to the filter inputs, and using an oscilloscope to measure the gain at a range of frequencies above and below the cutoff frequency in order to graph the response of the filter to different frequencies.

1.6.2.2.3 DATA ACQUISITION

Because of the high frequency of the signals we would like to measure, the data acquisition device must have a sampling rate of at least 80 MHz. Tektronix TDS 2024 oscilloscope will be used to gather data which will then be digitally filtered and analyzed using a LabVIEW program.
Testing of the Tektronix TDS 2024 will be accomplished with the use of a function generator. The outputs of the function generator will be connected to the inputs of the data acquisition device, and we will ensure that the function is correctly reconstructed with several different function parameters, including function shape, frequency, and amplitude.

1.6.2.2.4 DATA MANIPULATION

The digitized signal will not be inspected directly to differentiate between cell types, rather, useful information will be extracted from the signal and these data will be compared to previously recorded data to determine the cell type in a process called classification.

The TACS software system will employ digital filtering to increase the signal to noise ratio before signal processing takes place. We will determine the frequency ranges at which useful information is located by experimentation. Digital filtering will be implemented in Labview and will be tested by a visual comparison of the power spectrums of the input and the output of the filters.

Feature extraction is a process by which the information contained in a signal is used to create a feature vector, a list of properties and numerical values which represents a given signal. The feature vector will represent a number of different properties of the signal. The accuracy of the classifier increases based on the size of the input vector, assuming that each of the features, in a real-time system, the number of features must be limited to ensure fast computation time.

The feature vector will likely include power, variance, peak detection at different frequencies, covariance with a number of template waveforms, and wavelet spectrum matching with a number of different mother wavelets. Based on computational experiments, we will determine which features display the greatest quantitative difference between stained cells and unstained cells. Signal features will be extracted from the voltage responses of each of the PVDF films.

Testing of the feature vector will be accomplished by mathematically computing the values of a set of data and manually inspecting the feature vectors for a limited amount of input signals.

1.6.2.2.5 CLASSIFICATION

To differentiate between cells which are stained and unstained, the TACS will employ a software classifier. The classifier must be fast, accurate, allow sorting to multiple groups (as opposed to binary sorters), and it must be easy to train so that the TACS can be easily adapted to different sorting applications. The classifier which best meets these considerations is the artificial neural network (ANN). An ANN can be automatically configured using a set of sample data which thoroughly represents each of the output groups. Matlab has a built in toolbox called the Neural Network Toolbox which can be used to dynamically generate and calibrate an ANN.

An ANN functions similarly to an actual set of neurons in the body, but is a bit more versatile. Each neuron is modeled by a function with one or more inputs. Each input contributes in either an excitatory or inhibitory way, i.e., each input has either a positive or negative sign associated with it. A real neuron has an “all or nothing” response, in which the neuron propagates an action potential if the total
effect of all excitatory and inhibitory inputs is enough to depolarize the cell membrane to threshold. In the ANN, the response of the model neuron is more versatile. It can propagate an inhibitory signal, it can output a signal proportional to the product of multiple inputs, or it can be a simple threshold as in a normal neuron. Best of all, all these functions are created automatically by means of a machine learning algorithm. Machine learning is accomplished in an ANN by use of a large set of sample data and a back-propagation algorithm.

Figure 7 - Training an ANN

As shown in the figure above, a training experiment must first be performed in order to acquire sample data. In the case of TACS, calibration of the system begins with sorted cells. Each group of cells would be passed through the system separately in order to gather sample data. This data is then filtered and processed similarly to how it would be processed in actual use. The feature vector is inputted to the ANN, and the back-propagation algorithm takes changes the weights of the inputs to each neuron, and the equations involved to minimize the error between the output of the ANN and the expected output which is known at the start of the experiment.

The ANN is tested dynamically during calibration. Some of the sample data is used to configure the equations and the input weights whereas the rest of the data is used to test the accuracy of the ANN under each configuration. This ensures the accuracy of the ANN.

### 1.6.2.3 GATE MECHANISM

There exists a need to separate the particles into the appropriate bins. A gate will need to be designed to regulate the flow into appropriate bins. Since the overall apparatus is small, a gate that can switch between open and closed with electrical stimulation would be ideal.

### 1.6.2.3.1 ACTUATING MAGNETIC GEL MATERIAL

The magnetic gel with channel would sit on top of an electrically trigged magnet. A LabVIEW program will be used to send the input signal to the magnet. The magnetic field would attract the rod shaped ferrous oxide particles towards the magnet and cause the gel to collapse on the channel. This would then stop the flow of focusing fluid through the channel and direct the flow of the cells into the appropriate collection bins in accordance with the theory proposed pertaining to the H-gate. The concern
of cells contacting a magnetic gel is erroneous as only saline focusing fluid flows through the magnetic gate.

1.6.2.3.2 MAGNETIC GEL PREPARATION METHODS

A design for the actuating gate mechanism is the use of a magnetic gel. The Magnetic gel would be formed over a mold of a straight channel. The mold would be made using photolithography techniques. The gel would then be peeled from the mold and bonded to glass to create a working channel. The bonding to glass will be accomplished using a plasma treatment. The gel will be punched and two tubes attached. The first tube will be the input of the channel while the other channel will be at the output.

1.6.2.3.3 POSSIBLE CHOICES FOR TYPE OF GEL

The gel to be used is undecided and will be determined through experimentation. The first choice would be a NIPA hydrogel. This gel has been selected as it possesses a low viscosity and would be most likely to be manipulated by the magnetic field. If this gel cannot hold the shape of the channel or cannot be effectively bonded to glass then other types of gel will be experimented with. The next option would be an agarose gel which is a litter more rigid. The next progression in the same logic would be a PDMS gel. Each gel has their potential drawbacks. The NIPA gel may not hold a channel and could prove hard to work with if needed to be kept hydrated at all times. The agarose gel is made from cooling a liquid. This means that the agitation necessary to immerse the magnetic particles may inhibit the cross-linking of the agarose. It is known that PDMS can be effectively bonded to glass using a plasma treatment method. Since the PDMS is a viscous liquid before curing a maximum amount of magnetic particles will be able to me immersed by simple stirring. However, this gel may be difficult to manipulate with the magnet due to its rigid nature.

1.7 PROTOTYPE

1.7.1 FLOW OF CELLS

The first flow aspect to the prototype is the straight channel made out of PDMS. Cells flow through this channel from the holding bottle to the rest of the device. The straight channel will function as the excitation and signal recording portion of TACS. The laser diode shines directly above the channel at a location upstream from PVDF films which will eventually detect the photo-induced ultrasound signals. The straight microchannel is constructed from a PDMS channel mold that is plasma cleaned to a glass slide. Standard tubing was inserted and sealed at the inlet and outlet locations of the channel. For the final design the output channel would lead to the next stage of the system, the H-Gate.
The H-Gate is the central flow station where the flow is redirected to the correct bin. The H-Gate consists of a center sample input channel that the straight channel feeding the particles will be connected to. There are also two other input channels in which saline solution will be flowed to direct the particles in the specific direction. In order to regulate which redirection channel is activated at a given time, a magnetic straight channel gate will be used on both of the channels. Saline solution containers will be connected via standard tubing to a magnetic gate and then to the H-Gate. Our prototype H-Gate consists of the five different tube openings but it is not connected to any other pieces, rather, the two redirection input channels have standard tubing tubes with valves that will help mimic the different gates. By the use of the mechanical valves the H-Gate flow theory was tested. Unfortunately, the channels in the prototype may be too large to be able to generate non-turbulent flow. The next generation of the H-Gate should have a channel width in the micro range in order to better test this theory.

The straight magnetic channels use the same straight channel pattern as the sample input channel. The difference between the two designs is that magnetic particles were mixed into the PDMS in order to induce a magnetic field that will force the channel to close when induced. The last part of the flow for the TACS device is the output channels of the H-Gate. These two output channels will deliver the system flow to the specific bin.

Each piece of the fluidic subunit was tested individually. First the straight channel was attached to PVDF films so that it could be used to test the laser diode excitation and ultrasound detection. The H gate was tested as described and the description of the magnetic channel test can be found in section 1.7.3.

1.7.2 ELECTRONIC SYSTEM

The electronic system consists of three major devices: a computer with LabVIEW installed, a National Instruments PXI Hub with PXI 6251 and a Tektronix TDS 2024 Oscilloscope. Additional parts to the electronic system include two serial connectors, three PVDF sensors and the laser diode and driver.
The use of an external oscilloscope allowed greater signal resolution, and the option to automatically average signals provided a better signal to noise ratio that we would have accomplished using similar methods using National Instruments hardware for signal acquisition. The TDS 2024 provides a sampling rate of up to 2 GHz and signal resolution of 8 bits. This method also provides greater customizability when it comes to triggering. The leads of the oscilloscope are disconnected from the protoboard when not in use. Data can be saved to a flash memory card and transferred to the computer with a USB connector. To ensure that the oscilloscope is functioning as expected, a function generator can be connected to the scopes and used to verify that the test signal is recorded.

To begin using this device, PVDF sensors are attached to the specimen using a soft piece of PDMS. The PDMS sticks to the specimen and the sensors, causing them to be in direct contact with the specimen without requiring them to be bonded on to it. This allows the user to test the photoacoustic response of several specimens without having to inject them into the microchannel. If it is desirable to use the microchannel, the soft piece of PDMS can be used to put the sensors in contact with the bottom of the microchannel. PVDF sensors are connected to the protoboard via a serial connector to enable quick reconfiguration if desired.

To test the PVDF sensors, the sensors are attached to a beaker as described in the previous paragraph. A 1 MHz ultrasound transducer is attached to a BK Precision 4011A function generator which generates a square wave with a frequency of 1 MHz, a magnitude of 5 volts, DC offset adjusted to zero, and the duty cycle and CMOS options were disabled. The transducer is placed in the beaker so that plane of the transducer is parallel or close to parallel with the plane on which the PVDF sensors rest. Water is then poured into the beaker, which provides a medium through which the ultrasound wave can travel.

Figure 8 - PVDF Sensor Test Setup
Oscilloscope readings show that the 1 MHz PVDF film records the highest amplitude response, and the other two sensors record lesser responses because the signal lies farther from their center frequencies.

The protoboard is connected to the PXI 6251 via a BMC 2120 socket board. Wires connect to channel CTR 0, the digital ground, and a 5 volt voltage source in order to drive the laser diode and trigger the oscilloscope.

The trigger for the laser diode has an amplitude of 5 volts, a pulse width of 150 nanoseconds, and a period of 1 millisecond. This is accomplished with the LabVIEW data acquisition express VI, which is incorporated into the TACSMain VI, which must be running to drive the laser diode. The same signal that drives the laser diode is used to trigger the oscilloscope. The oscilloscope records about 1 microsecond of data, beginning right before the laser diode is fires. The displayed data is the average of 128 repetition rounds.
The ETX 10A laser diode driver is sensitive to static discharge and is disconnected when not in use and stored in a protective antistatic bag. For easy removal and reconnection, the laser diode driver is attached to the protoboard via a serial connector.

To test the laser diode an LG VX5500 camera phone was used. This device is known to record signals in the near infrared range. The camera phone was positioned so that the laser was directed straight into the camera lens. If the laser does not point directly into the lens, the camera will not be able to record the light. Once the camera is correctly positioned, the laser diode driver is turned on and off using the LabVIEW program.
1.7.3 ACTUATING MAGNETIC PDMS GATE

The final choice for the magnetic gate was a PDMS gel. Many different types of gel were experimented with and found to have crucial flaws. The original suggestion of the client was to use a NIPA gel. This gel was found to soft to hold the shape of a channel when formed on a mold. An agarose gel was then experimented with. This gel, although easy to make, was not able to be effectively bonded to glass to create a working channel.

The PDMS gel was found to be especially advantageous in key areas. First, the extremely viscous nature of the PDMS monomer before polymerization allows for a maximum amount of magnetic particles to be suspended. These particles to not immediate collect at the bottom of the gel. Instead, with slight mixing, the particles are able to be uniformly dispersed. This allows for a consistent and predictable response of the gel to the magnetic field.

Another advantage of the PDMS gel was its ability to be bonded to glass. Since PDMS has been repeatedly used for other microchannel devices in research, there were available protocols and machinery necessary to create a working channel.

The PDMS in the prototype was made with the sylgard® 184 silicone elastomer kit. The common mixing ratio for monomer to curing agent is 10:1 by weight. However, through experimentation, it was found that gels with varying rigidity would be made by changing the common ratio. The ratio used in the prototype to achieve the softest gel that would still cure was 40:1 monomer to curing agent.

The amount of magnetic particles added was also determined through experimentation. The highest weight percent of particles that could be added to a gel that cured was used. Thus, the final weight ratios of monomer: magnetic particles: curing agent used to make the prototype gel was 40:1:33.

The gel formed over a mold of a straight channel. This leaves an imprint of a channel on in the magnetic gel. This gel was then bonded to glass using a plasma treatment method with the imprint of the channel facing the glass. The ends of the channel were punched and tubing was attached. The prototype of the gate then consists of a working channel made up of a magnetic PDMS gel.
2 REALISTIC CONSTRAINTS

Any design project has certain engineering standards and constraints that need to be considered before and during production. Only if the standards and constraints are met can the design be put into production. The TACS device is no exception. Great precision will be needed when designing and assembling the device. The final product needs to operate effectively and safely yet hopefully have a low cost. Since there are three different parts to this device, the flow system, electronics system, and the materials system, they need to function independently and more importantly they must all work in conjunction. Certain engineering standards need to be met in order for these three systems to function properly together.

2.1 ECONOMIC CONSTRAINTS

There are a few economic constraints to this device. Although given a projected budget of $5,000 there are hopes to spend much less money than that. Due to the research based aspect of this project there will be many trial and error experiments fine-tuning our device or even simply seeing if these methods can be applied to cell sorting. There is a small chance that this device may not be able to be used for cell sorting and the money gone towards this cause could have gone to waste. Unlike other projects, our group is simply testing a theory and trying to apply it to the fields of flow cytometry and cell sorting.

2.2 ENVIRONMENTAL CONSTRAINTS

Another big constraint to this device will be the surroundings, or environment in which it is used. Due to the sensitivity of the PVDF films to detect sound, it may be required to use this device in as quiet a setting as possible. Any sort of vibration, wind, or even talking could affect the PVDF wave detection. In order to overcome this, it may be necessary to use a sound-proof casing that will cover the device, either entirely or just over the PVDF film and capillary tube section. This will hopefully drown out the sound that could potentially be picked up by the system.

2.3 SUSTAINABILITY

If this device needed to get disposed of it will hopefully not contain any materials or components that will harm the environment. With current efforts to limit harmful chemicals polluting the atmosphere it is only fair if this was taken into consideration when designing this device. As of now we have not been faced with such a decision.

2.4 MANUFACTURING CONSTRAINTS

TACS contains a lot of small and brittle components such as the capillary tubing, PVDF films, laser diode, and the Zeolite crystal. When manufacturing components of this device and putting it together
great care will need needed to ensure proper assembly. Since the capillary tubes are glass and very thin, handling these with care will be a priority in the design lab. Also, the PVDF films are extremely thin and may rip under small forces. Since the PVDF films have already been treated to obtain certain molecular properties it cannot be welded. In order to overcome this during assembly conductive epoxy will be used to attach this to the capillary tubes. Another manufacturing constraint is the application is the Zeolite crystals, these crystals require rigorous effort in order to be created. There may not be a facility at UConn that can host such a procedure.

2.5 ETHICAL CONSTRAINTS

There may be ethical concerns to this device. Future uses of this device include being implanted inside the human body for drug delivery, blood filtration, or tissue removal. If this device were to be implanted in the lymphatic system another exciting potential use of this device could be to filter cancerous cells. Some people may find these uses unnatural and could potentially look down upon this device. In the first stages of this device the future uses will not be considered, but simply trying to apply different methods to cell sorting.

2.6 HEALTH AND SAFETY CONSTRAINTS

This device contains a few potentially health and safety hazards. The live voltages applied to the actuating material need to be contained in a safe way to avoid user shock when either operating the device or assembling the device. Another potential hazard during assembly is the handling of fragile and thin glass capillary tubes. If one of the tubes were to break the probability of puncturing skin greatly increases.

2.7 SOCIAL AND POLITICAL CONSTRAINTS

Just like ethical concerns, social and political concerns may arise later if this device is used or implanted in vivo. Since this may not seem natural to some people and there are groups who disagree with the production.
SAFETY ISSUES

With the realization and testing of the TACS there are certain safety concerns that must be considered.

There will be many electrical components in this device. And the device will contain an electrolyte solution which conducts electricity better than pure water. There will be connections to the PVDF films as well as to the actuating material. There also will be outputs from the computer and National Instruments device. It is then important to be aware when components are turned on and that there can be a risk of electrical shock. This is a concern, but only a small voltage (approximately 500 mV) will be used for output to the gates. Electrically insulating gloves may need to be worn during testing and setup since there are unknown factors that could lead to erratic responses. Care will have to be taken to ensure that no solution from the tanks is leaked and all the electrical components remain dry. The capacitors on the laser diode driver charge to voltages exceeding 200 Volts. Extreme care will be taken to protect the operator during device use as well as team members during device construction.

Fabrication of the electrically actuated gates will involve a number of hazardous chemicals and materials. Purification of the zeolite actuating gate material and the use of colored particles or cells require certain laboratory processes and chemicals. For this part of the project, common full chemical laboratory safety practices must be employed. This will include the use of protective wear such as gloves, goggles, closed-toed shoes, and work coats. Additionally, environmentally conscious disposal procedures will be practiced. This includes the proper disposal of any organic or inorganic wastes and a special care in the case of bio-hazardous or carcinogenic chemicals.

Any hazardous chemicals used in the photolithography process will be used under a fume hood to prevent accidental inhalation. Brittle silicon wafers will be handled with utmost care as they can shatter into shards.

Appropriate warnings will be provided in the TACS Operator’s Manual as well as with labels on the device.
4 IMPACT OF ENGINEERING SOLUTIONS

Completion and implementation of this proposed optimal design may potentially have a positive impact globally on humanity and the environment.

As stated above, the eventual objective is to fabricate a cell sorting device that can be adapted for in vivo use. This could result in many health benefits in the future. This device could allow for the filtration of targeted cells. The most likely location for the implantation of this device will be the circulatory system. If implanted in the circulatory system, the device could potentially remove harmful cells and particles from the bloodstream. The technology to tag certain cells based on protein synthesis exists. The in vivo cell sorter could remove these targeted cells. The sorter may help those with compromised immune systems get rid of infectious or foreign cells.

Additionally, the ability to sort certain targeted cells could allow for the administration of drugs to targeted cells, rather than allowing diffusion of the drugs into all cells including those which would be unaffected or damaged by the drug. Many patients shy away at the thought of taking medication due to the risk of unwanted side effects. This application would reduce those side effects and could increase the number of patients willing to undergo necessary drug treatment.

The device could also be used to sequester a sample of specific cells from the blood for further investigation, without having to remove an entire blood sample then separate the cells. If implanted in the lymphatic system, the filtration and isolation of cancerous cells would be the final goal.

In addition to the inherent health benefits, the in vivo cell sorter can also make a global economic and environmental impact. This device may reduce the need for complicated laboratory testing and sorting of cells. Laboratory methods require skilled, trained and paid operating personnel as well as expensive and bulky equipment. In addition to being a more cost effective solution, the in vivo cell sorter could offer a more environmentally sound option versus the current laboratory methods. The manufacturing and transportation of bulky laboratory equipment can increase the amount of dangerous gases in the air that are harmful to the human respiratory system and also the environment though global warming and ozone layer depletion. Laboratory methods also generate waste products that further pollute the environment.
5  LIFE LONG LEARNING

As a result of working on this project, we the members of the TACS design team have learned and will continue to learn engineering, business, and life skills.

The engineering skills learned include increased experience with specialized software tools including Autodesk Inventor, National Instruments Multisim, National Instruments LabVIEW Integrated Development Environment, Microsoft Visio, and Matlab especially the Neural Network Toolkit. We have also and will continue to gain experience and knowledge in the fields of piezoelectric and nanoporous materials including the preparation, testing, and field response of such materials. We will also gain knowledge and experience in implementing the thermoacoustic and photoacoustic principles.

Our goal is to prove that the theories we are testing will work in a micro sized device, and hopefully help Dr. Kotha to obtain a patent for a device which implements these theories. This is very important from a business perspective, as it ensures that the credit for this device remains at the University of Connecticut, and that Dr. Kotha can possibly receive a grant to develop this device.

We are striving to produce a device which is efficient, fast, and accurate, while being as cost-effective as possible. Although we have a much larger budget than we will need, we do not want to spend any money that doesn’t need to be spent. Research for procurement of the most cost-effective components is a valuable skill in the business world, as it leads to being able to produce the highest quality at the lowest cost to customers. It is also a skill which will help us to make informed purchasing decisions on a day-to-day basis.

The TACS project integrates a variety of different engineering concepts in order to create a useful device. To our benefit, the members of the team have varied interests and are able to split the project into parts which each of us finds interesting. As a group, we had all had relatively limited interaction with each other. This proves to be a challenge because we will need to accommodate for the learning and working styles of our teammates. We will need to practice proper communication, tolerance and cordiality towards team members and others involved as well as a sense of responsibility to uphold an individual commitment to the project.

We are learning to work as a team and to delegate responsibilities evenly amongst group members. We want to minimize the amount of overlap between our assignments and increase efficiency on a personal level as well as on a group level.
## 6 BUDGET

### 6.1 BUDGET

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7 TEAM MEMBERS CONTRIBUTIONS TO THE PROJECT

The TACS team is composed of three biomedical engineering students at the University of Connecticut. Each has a different specialization within this program and will contribute a different skill set and point of view to the project.

8.1 MICHAEL LALLI

Michael is the completing the bioinformatics track and thus concerns himself with the software and electronic portions of the device.

He is responsible for the procurement of the PVDF films, conductive epoxy, and the laser diode and its associated electronics, as well as a dye which will absorb the frequency of the laser diode. He is responsible for the circuitry associated with acquiring the photoacoustic signal from the PVDF films using the NI DAQ board and Oscilloscope. He will also be responsible for the circuitry associated with triggering the laser diode, and actuating the microfluidic valves. These responsibilities include preparing the PVDF films by cementing leads to them with the conductive epoxy, and testing their resistance and performance.

Michael has performed several experiments to determine the most effective configuration to acquire a signal using PFVD films and the optimal pulse width and pulse energy of the laser diode to produce photoacoustic waves.

Michael has designed the software architecture necessary to analyze the data and classify test particles.

1.2 NIMESH PATEL

Nimesh’s concentration is in biomaterials, thus it was natural for him to take on the objectives of developing the micro fluidic valve.

He found a commercially available actuating valve to use in the project from the Dolomite Company. This was ideal because the gate actuates in a very timely fashion (8ms), and come with simple connections and instructions for use. This valve employs a relatively low voltage (12V) which leads to improved safety of the product. Additionally, it is an in stock item and can be shipped out of the Massachusetts branch in 24 hours. However it was found that this was not suitable for the client since it was not novel.

Despite finding a suitable actuating fluid valve, efforts were focused on the development of the client’s original suggestion of zeolite. Since there was no background to build on this method proved too difficult. Research was then conducted on electrically expanding hydrogels. After finding that these would not provide the necessary volume changes, Nimesh focused on making a magnetic gel.

He researched and developed three different types of gels. Nimesh then found that the PDMS gel would be the most viable option due its ability to be bonded to glass and to hold the most magnetic
particles. He then fabricated a working magnetic gel channel. He then tested the prototype and found that it was able to slow the flow through the channel.

7.1 DAVID PILLITTERE

David’s track is biomechanics, so for this project he is primarily responsible for modeling and designing the system through with cells and supporting saline will flow. He produced CAD drawings of any of the system’s microchannels and flow system using Autodesk Inventor. He was also responsible for researching the mathematics behind the photoacoustic theory, and working with Microsoft Project to map out the semester timelines. Determining the sequence of flow was a difficult task for this project because the cellular flow essentially is the common point for all of the components of the device. The cells needed to have the correct exposure for each stage of the device. One specific part of the flow will be the cell excitation ultrasonic wave detection through the microchannels. The rate of the flow through the microchannels needed to be known so the correct time delay can be integrated in the peripheral computing system. Part of the semester David was responsible for researching spin coating machines for purchase for Dr. Kotha’s lab.

The majority of David’s time for this project has been spend on fabricating various microchannel designs in order to find the most efficient and applicable method. In order to fabricate microchannels the use of a spin coater is needed to ensure accurate and precise dimensions. Plasma treating the various microchannels to glass was the next step for microchannel fabrication. David gained a lot of exposure to different laboratory equipment and procedures throughout the year. David was successful in creating a straight magnetic gate channel, a regular straight channel for flow input, and an H-Gate used to direct the cellular flow. The different channels incorporated the use of a syringe in order to test the flow mechanics.
CONCLUSIONS

The goal of this project was to determine the feasibility of decreasing the cost of flow cytometers by implementing a range of cost-saving measures. These cost-saving measures include the use of a laser diode for cell excitation rather than the more commonly used and much more expensive CO₂ and crystal lasers, the use of PVDF films to detect ultrasound rather than commercial ultrasound transducers, and the use of electrically actuated microgates, which consume far less materials and would be more cost effective than the electromagnetic devices used in current flow cytometers.

A laser diode can be used for cell excitation; however the least expensive laser diodes are in the infrared range. This raises an important issue because biocompatible dyes in this range are much more expensive, as their use is less common and less researched than dyes in the visible range. Although this type of system could save money on upfront costs, it would raise the price of using the device. However if the system could be configured to use a lower power laser diode in the visible range such as those used for burning compact discs, the price of the device and its associated dyes could be greatly reduced.

Unfortunately the PVDF films did not provide a high enough signal to noise ratio for us to see the photoacoustic signal from our selected dye. The problems are that the sensors are too sensitive to the electrical noise from the trigger signal, and that the photoacoustic signal by definition is miniscule. Further filtering and signal analysis could perhaps yield better results, but will be difficult given how small the signals are.

The use of an H-gate design for the control of fluid allows for the cells themselves not to come in contact with the gate or the magnetic particles. This H-gate design effectively routes the cells to the appropriate collection bin while ensuring that a cell will never be closed upon by the gate.

The use of a PDMS microchannel to control the device flow proved optimal. Fluid was able to be injected through the channel without spilling out the sides. The PDMS channels are time consuming but easy to fabricate allowing for mass production if called for.

An electrically actuated gate allows for an easy and fast responding micro-fluidic valve. The use of a PDMS magnetic gel makes this portion of the device bio-compatible. The materials for this magnetic gel gate are also inexpensive and easy to assemble.

However, it was found the PDMS proves difficult to move with a magnetic field due to its rigidity. This rigid PDMS gel was shown to slow the fluid flow but not to stop it entirely. This problem may be solved in the future by manipulating mixing ratios to make a softer and more malleable PDMS magnetic gel gate. Another solution may be to find a method to bond a softer gel, such as an NIPA gel, to glass to create a working channel.

There is a range of different applications for such a device from the collection of desired test cells to the filtration of cancerous cells or other harmful particles. A working prototype of this device would prove useful for the medical community, by cutting costs and speeding diagnostic results.
9 WORKS CITED


10 ACKNOWLEDGEMENTS

Carol Norris is the faculty scientist for flow cytometry and electron microscopy at the University of Connecticut campus at Storrs. After touring her laboratory, our group approached her to learn more about the equipment. Carol provided us with CaliBRITE test particles to use in calibrating the device. This was a big contribution from her and our group greatly appreciated it. The CaliBRITE bead set contains four samples of particles. There are three tubes which contain different colored 6 micron diameter test beads, and a tube of uncolored control beads.

Jinzi Deng is a PhD student under Dr. Shor in UConn’s Chemical Engineering department. She has been a resource for our group throughout the second semester of the senior design process. Jinzi helped train David on the spin coating machine and helped train both David and Nimesh on the plasma cleaner. She has given up some of her spare time to oversee the use of the equipment in the lab. Our group would like to acknowledge and thank Dr. Shor and Jinzi for their continuous cooperation and for allowing us to use their facilities. Our project would have not reached the point it is at now without their kindness.

Sterling Nesbit was an enormous help on this project. He is a PhD student under Dr. Kotha, and was willing to supervise us in the spin coating laboratory as well as provide insight regarding ultrasound generation and acquisition. He was very approachable and was a great resource when we had questions regarding hardware.

Quinn Zhu, a professor in the Electrical and Computer Engineering department was kind enough to meet with Michael and discuss the different options for laser diodes and ultrasound sensors.

Katelyn Burkhart, an undergraduate student in Bio-medical engineering working for Dr. Kotha, showed Nimesh how to make NIPA gels

Yamalia Roberts, a PhD student, made magnetic nano-particles for the team to use in the device.

Dr. Shiva Kotha, a Bio-medical Engineering professor at the University of Connecticut, helped the group with much insight and many resources that allowed the project to progress. Not only did he fund the project, but provided a helping hand and guided our path as we developed this device.
## APPENDIX

### UPDATED SPECIFICATIONS

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Cell Processing Rate</strong></td>
<td>&gt;6 cells/min</td>
</tr>
<tr>
<td><strong>Microchannels</strong></td>
<td></td>
</tr>
<tr>
<td><strong>Straight Channels</strong></td>
<td></td>
</tr>
<tr>
<td>Length</td>
<td>55.5 mm</td>
</tr>
<tr>
<td>Width</td>
<td>3.0 mm</td>
</tr>
<tr>
<td>Height</td>
<td>500 μm</td>
</tr>
<tr>
<td><strong>H-Gate</strong></td>
<td></td>
</tr>
<tr>
<td>Input Channel Length</td>
<td>28 mm</td>
</tr>
<tr>
<td>Side Channel Length (x4)</td>
<td>7.5 mm</td>
</tr>
<tr>
<td>Width</td>
<td>3.0 mm</td>
</tr>
<tr>
<td>Height</td>
<td>500 μm</td>
</tr>
<tr>
<td><strong>Electrical</strong></td>
<td></td>
</tr>
<tr>
<td>Input Voltage</td>
<td>5 V</td>
</tr>
<tr>
<td>TTL Modulation Voltage Output (High)</td>
<td>3.3 V &lt; X &lt; 6 V</td>
</tr>
<tr>
<td>TTL Modulation Voltage Output (Low)</td>
<td>&lt; 0.8 V</td>
</tr>
<tr>
<td><strong>Laser Diode</strong></td>
<td></td>
</tr>
<tr>
<td>Peak Power Output</td>
<td>25 W</td>
</tr>
<tr>
<td>Wavelength</td>
<td>850 nm</td>
</tr>
<tr>
<td>Pulse Width</td>
<td>1-100 ns</td>
</tr>
<tr>
<td>Emitting Area</td>
<td>200μm × 2μm</td>
</tr>
<tr>
<td>Duty Cycle</td>
<td>0.1 %</td>
</tr>
<tr>
<td>Operating Temperature</td>
<td>-40 – 85°C</td>
</tr>
</tbody>
</table>
Fluid

Composition for First Round of Experiments  Water
Operating Temperature  18-25°C

PVDF Films

Thickness  28, 52, and 110 μm

Data Acquisition

Sampling Frequency  2 GHz
Resolution  8 bit

Software

For each specimen that passes through the system, the system must enervate the specimen multiple times according to the optimal value found by experiment. For each enervation, the voltage signal from each of the PVDF films is sampled at a rate of 1GHz, and the data is recorded and saved. The software then averages this data together, and programmatically determines which group the specimen fits, then outputs an analog voltage to the correct microfluidic gate.
11.1 DATA SHEETS

11.1.1 ETX-10A LASER DIODE DRIVER DATA SHEET

ELECTRO-OPTIC DEVICES, INC.

ETX-10A
Single Supply, 5-Volt, Pulsed Laser Diode Driver
With Integrated HV Power Supply

FEATURES:
- Compact pulsed laser diode driver design.
- Integrated high voltage power supply.
- Single 5V supply requirements (3.3V also available).
- Pulse widths from 5 to 30 nanoseconds (typically).
- PRF up to 10 kHz with integrated power supply. (see specifications)
- PRF in excess of 50 kHz with external HV power supply (eg: model EHV-4)
- Discharge current up to 75 Amps (pk). (see specifications)
- Peak discharge current may be set locally via trim-pot or by external control voltage (XCV).
- Pulse width can be modified via selection of capacitance and resistance banks.
- Universal laser mounting pads accommodate most 2 and 3 lead pulsed laser diodes such as all OSRAM (Infineon) SPL series diodes and Perkin-Elmer "M", "U" and "W" pkgs.
- ETX-10A Laser Drivers are available with OSRAM SPL series lasers installed with test data. Optional: extended test data including Po vs. Vds plot and laser emission spectral plot.
- Orthogonal mount of laser diode provides minimum inductance. Facilitates a complete Faraday cage for reduced EMI when a metal shield (available separately) is attached to the PCB's perimeter ground plane.
- Compact PCB just 2.25 x 1.125 inches.

DESCRIPTION:
The ETX-10A is a compact pulsed laser diode driver with an integrated 0-200V high voltage switching power supply. Integrating the high voltage power supply into this pulsed laser driver, and eliminating the need for an external high voltage supplies, allows the ETX-10A to be powered by a single 5V power supply. This minimizes extra design work, board space, and cost of an external power supply by keeping the required high voltage supply needs on-board.

A modifiable pulse width gives a configurable PRF up to 10kHz (integrated power supply) or in excess of 50kHz (High performance external power supply: eg. model EHV-4).

Minimizing pin-out, control is obtained by three control signals: XCV, SHDN and TRIG. XCV enables ability to exchange input/auxiliary discharge voltage, which can also be done by trim-pot VR-1, and will indirectly modify the peak discharge current. SHDN shuts down the ETX-10A to enter a power saving mode that draws only 15mA of current from the supply. TRIG triggers a high voltage pulse, which in turn can be monitored by the discharge monitor (DM) pin for more accurate measurements.

Flexible design layout provides accommodates most 2 and 3 lead pulse laser diodes. Orthogonal mount of laser diode provides minimum inductance that, when a metal shield (available separately) is attached to the PCB's perimeter ground, will facilitate a complete Faraday cage for reduced EMI. PCB is very compact with board dimensions of 2.25 x 1.125 inches.

ETX-10A pulsed laser diode drivers are available with OSRAM SPL series lasers installed and tested. Package is also provided with test data and optional extended test data that includes Po vs. Vds plot and laser emissions spectral plot.
PI Connections:

The ETX-10A implements an eight-conductor interface (PI) using 1mm pitch flat flex cable. The ETX-10A is available with or without the 8 pin 1mm FFC connector installed (see ordering codes). The pin connections are as follows:

<table>
<thead>
<tr>
<th>Pin</th>
<th>Signal</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>DM</td>
<td>Discharge Monitor Output</td>
</tr>
<tr>
<td>2</td>
<td>GND</td>
<td>Ground</td>
</tr>
<tr>
<td>3</td>
<td>SHDN</td>
<td>Shutdown</td>
</tr>
<tr>
<td>4</td>
<td>GND</td>
<td>Ground</td>
</tr>
<tr>
<td>5</td>
<td>XCV</td>
<td>External Control Voltage</td>
</tr>
<tr>
<td>6</td>
<td>+P+</td>
<td>Power Supply</td>
</tr>
<tr>
<td>7</td>
<td>N.C.</td>
<td>No Connection</td>
</tr>
<tr>
<td>8</td>
<td>TRIG</td>
<td>Trigger Input</td>
</tr>
</tbody>
</table>

Vext. – External Discharge Voltage Input / Test Point:

<table>
<thead>
<tr>
<th>Pin</th>
<th>Signal</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>GND</td>
<td>Ground</td>
</tr>
<tr>
<td>2</td>
<td>Vdischarge</td>
<td>Discharge Voltage (0 – 200 Vdc)</td>
</tr>
</tbody>
</table>

Signal Details:

DM, Discharge Monitor Output – This output has a negative going rapid transition that is temporally coincident with the laser pulse emission. In laser ranging applications, this signal can be considered time zero (0) and is normally viewed by a fast voltage comparator in the user's system.

TRIG, Trigger Input – This CMOS/TTL compatible input triggers the ETX-10A to fire the laser diode. The driving circuit should be capable of sourcing a trigger pulse of at least 3.5Vdc peak and at least 50ns width. High repetition rate applications should make certain to limit the trigger pulse width as the discharge capacitor bank cannot begin to recharge while the trigger input remains logic HI.

SHDN, Shutdown – This CMOS/TTL compatible input disables the ETX-10A switching supply while in the logic HI state. The Shutdown (SHDN) input can be used to minimize current consumption and gate switching noise during sensitive operations (see AN-3, A Basic Time-of-Flight Pulsed Ranging System).

XCV, External Control Voltage – The External Control Voltage (XCV) analog input controls the laser discharge voltage. The XCV input can be varied over the range 5V – 3.5Vdc to proportionately set the laser discharge voltage over the range of 0V to 200Vdc.

Universal Laser Diode Mount:

The laser diode mounts to the bottom of the ETX-10A (the side opposite components). Be certain to match Anode (A) and Cathode (K) connections according to laser diode specifications. Failure to do so can result in damage to the laser diode. The ETX-10A features a universal laser diode pad configuration. The pad spacing and hole sizes accommodate most popular laser diode packages including all OSRAM SFL series laser diodes and Perkin-Elmer’s “R,” “S,” “U” and “W” style packages. In each case, the hole openings are such that the diode can be installed with the laser emitting area centered on the ETX-10A mounting holes. For plastic encapsulated lasers (OSRAM SFL and P-E type “W”) which have the laser diode offset from the centerline of the leads, the outer laser mounting holes on the ETX-10A are large enough to allow the laser diode to be counter-offset by approximately 0.2 mm to allow better centration on the ETX-10A mounting holes.

CAUTION: METAL PACKAGED LASER DIODES ARE CHARGED TO VDISCHARGE POTENTIAL! Care must be taken to avoid shock hazard and to prevent shorting contact to metal mounting features.

Local Adjustment of the Laser Discharge Voltage:

The laser discharge voltage, and correspondingly the peak laser current, can be adjusted at trim pot VR1. The discharge voltage is variable over the range of 0 to 200Vdc.

Modifying Peak Current and Pulse Width:

The peak current discharged through the laser is proportional to the laser discharge voltage (the potential across the capacitors) and the total capacitance of the capacitor bank. To increase the peak current through the laser, the discharge voltage should be increased. This will, however, increase the pulse width slightly.

Pulse width can be modified by choice of total discharge capacitance and effective series resistance. Capacitance as low as a couple hundred picofarads can develop very short and intense laser pulses when minimal series resistance (Rbank) is used.

The standard configuration for the ETX-10A provides a capacitance bank of 4nF (total) and a series resistance of 0.55 Ohms. This selection provides a nominal pulse width near 15ns at a discharge voltage of 50Vdc. The ETX-10A can accommodate up to five discharge capacitors.

Recommended VDischarge Operating Range:

Operation at low discharge voltages leads to increased sensitivity to pulse width variations. Depending upon the installed laser diode, discharge voltages less than 20 – 40 Volts should be avoided. If lower laser output power is desired, use less discharge capacitance in the banks to allow a higher discharge voltage.
ETX-10 OPERATING SPECIFICATIONS:

<table>
<thead>
<tr>
<th>PARAMETER</th>
<th>MIN.</th>
<th>TYP.</th>
<th>MAX.</th>
<th>UNIT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Supply Voltage (Vdc)</td>
<td>4.8</td>
<td>5.0</td>
<td>5.2</td>
<td>V dc</td>
</tr>
<tr>
<td>Supply Current (Icc)~</td>
<td>15</td>
<td>25</td>
<td>60</td>
<td>mA dc</td>
</tr>
<tr>
<td>Supply Current during shutdown (Issh)</td>
<td>15</td>
<td>20</td>
<td>200</td>
<td>V dc</td>
</tr>
<tr>
<td>Laser Discharge Voltage (Vdls)</td>
<td>5</td>
<td>50</td>
<td>200</td>
<td>V dc</td>
</tr>
<tr>
<td>Pulse Repetition Rate (continuous)~</td>
<td>50000</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulse Repetition Rate (contin. ipk=20A,Vdls=35V)~</td>
<td>6000</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulse Repetition Rate (contin. ipk=30A,Vdls=45V)~</td>
<td>2500</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulse Repetition Rate (contin. ipk=40A,Vdls=115V)~</td>
<td>1200</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulse Repetition Rate (contin. ipk=50A,Vdls=145V)~</td>
<td>700</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulse Repetition Rate (contin. ipk=60A,Vdls=180V)~</td>
<td>400</td>
<td>Hz</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Maximum Current Pulse Amplitude (peak)~</td>
<td>60</td>
<td>75</td>
<td>15</td>
<td>A</td>
</tr>
<tr>
<td>Current Pulse Width (at 50% amplitude)~</td>
<td>5</td>
<td>15</td>
<td>30</td>
<td>ns</td>
</tr>
<tr>
<td>Trigger Pulse Amplitude (peak)</td>
<td>3</td>
<td>5</td>
<td>7</td>
<td>Vdc</td>
</tr>
<tr>
<td>Trigger Pulse Width (at 50% amplitude)~</td>
<td>30</td>
<td>50</td>
<td>200</td>
<td>ns</td>
</tr>
<tr>
<td>Shutdown Time (lsw)~</td>
<td>5</td>
<td></td>
<td>5</td>
<td>μs</td>
</tr>
</tbody>
</table>

NOTES:
1. Dependent upon repetition rate, discharge voltage and capacitance bank.
2. May require external laser power supply - dependent upon peak current, pulse width and repetition rate. NOTE: Excess heat dissipation may damage capacitor bank or laser diode. Do not exceed 125mW average dissipation per discharge capacitor without forced cooling.
3. Measured load: * length of 100A, rise with current probe attached. Vdls measured at Vdc. test point using internal laser power supply @ 25°C, pulsewidth typically 15μs @ Vds = 100Vdc for standard ETX-10A configuration. 4mF capacitor bank + 0.55Ω series resistance.
4. Dependent upon laser package inductance, discharge capacitance, series resistance and laser voltage.
5. Time it takes the ETX-10A to enter shutdown mode from the instant the SHDN pin becomes high.

ETX-10A Ordering Options:

ORDERING CODES:

<table>
<thead>
<tr>
<th>CODE</th>
<th>DESCRIPTION</th>
</tr>
</thead>
<tbody>
<tr>
<td>ETX-10A</td>
<td>ETX-10A laser driver without 6 position connector, no Laser</td>
</tr>
<tr>
<td>ETX-10A-C</td>
<td>ETX-10A laser driver with 8 position FFC connector, no Laser</td>
</tr>
<tr>
<td>ETX-10A-65</td>
<td>ETX-10A laser driver, no connector, SPL PL65 Laser Diode*</td>
</tr>
<tr>
<td>ETX-10A-90</td>
<td>ETX-10A laser driver, no connector, SPL PL90 Laser Diode*</td>
</tr>
<tr>
<td>ETX-10A-93</td>
<td>ETX-10A laser driver, no connector, SPL PL90_3 Laser Diode*</td>
</tr>
<tr>
<td>ETX-10A-85-C</td>
<td>ETX-10A laser driver, FFC connector, SPL PL65 Laser Diode*</td>
</tr>
<tr>
<td>ETX-10A-90-C</td>
<td>ETX-10A laser driver, FFC connector, SPL PL90 Laser Diode*</td>
</tr>
<tr>
<td>ETX-10A-93-C</td>
<td>ETX-10A laser driver, FFC connector, SPL PL90_3 Laser Diode*</td>
</tr>
<tr>
<td>*</td>
<td>Includes test data indicating measured peak laser power at Vdls = 50Vdc</td>
</tr>
</tbody>
</table>

Accessories:

<table>
<thead>
<tr>
<th>CODE</th>
<th>DESCRIPTION</th>
</tr>
</thead>
<tbody>
<tr>
<td>ETX-SHIELD</td>
<td>Brass EMI shield to fit ETX-10A and ETX-nX series laser drivers</td>
</tr>
<tr>
<td>ETX-FFC-CON</td>
<td>8 Position, 1mm FFC Connector, Through-hole</td>
</tr>
<tr>
<td>ETX-FFC-RIB</td>
<td>8 Conductor, 4&quot; length, 1mm FFC Ribbon Cable</td>
</tr>
<tr>
<td>ETX-10A-ETO</td>
<td>Extended test data including Po vs. Vdls and laser spectral plots</td>
</tr>
<tr>
<td>ETX-10A-NRE</td>
<td>Custom pulsewidth / power output Non-Recurring Engineering</td>
</tr>
</tbody>
</table>
ETX-10A Mechanical Dimensions:

![Diagram of ETX-10A mechanical dimensions]
11.1.2 SPL PL85 LASER DIODE

Nanostack impuls-Laserdiode im Plastikgehäuse, 10 W Spitzenleistung
Nanostack Pulsed Laser Diode in Plastic Package, 10 W Peak Power

Lead (Pb) Free Product - RoHS Compliant

SPL PL85

**Besondere Merkmale**
- Optische Spitzenleistung bis zu 10 W
- Laserwellenlänge 850 nm
- Geeignet für kurze Laserpulse von 1 bis 100 ns
- Austrittsoffnung 200 µm x 2 µm
- Kostengünstiges Plastikgehäuse für hochvolumentige Anwendungen

**Features**
- Optical peak power up to 10 W
- Laser wavelength 850 nm
- Suited for short laser pulses from 1 to 100 ns
- Laser aperture 200 µm x 2 µm
- Cost effective plastic package for high volume applications

**Anwendungen**
- Tragbare LaserentfernungsMESSgeräte für Golf, Jäger, Bauingenieure
- Automobilanwendungen (Intelligenter Tempomat, Aufbraul-Vorwarnkennung, Kollisionsvermeidung, adaptive Rückleuchten)
- Verkehrsüberwachung (Laserradarpistole, Verkehrszählung, Fahrzeug-Klassifizierung, Abstandsmeßung, Nebelkennung)
- Professionelle Lasersensoren für Abstandsmeßung, Positionierung, Sicherheit

**Applications**
- Hand-held Laser Range Finders (LRF) for golfers, hunters, civil engineers
- Automotive applications (Adaptive Cruise Control (ACC), pre-crash detection, collision avoidance, adaptive rear lighting)
- Traffic surveillance (Laser speed gun, traffic recording, vehicle classification, distance measurement, fog detection)
- Professional laser sensors for distance measuring, positioning, protection

**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-Norm 60825-1 behandelt werden.

**Safety Advises**
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 “Safety of laser products”.

2009-03-04

Opto Semiconductors

OSRAM
### Grenzwerte (kurzzeitiger Betrieb) (Umbgebungstemperatur $T_{ah} = 25 \degree C$)

Maximum Ratings (short time operation) (Ambient temperature $T_{ah} = 25 \degree C$)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Werte Values</th>
<th>Einheit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Spitzenausgangsleistung</td>
<td>$P_{peak}$</td>
<td>13</td>
<td>W</td>
</tr>
<tr>
<td>Peak output power</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spitzendurchlaßstrom</td>
<td>$I_F$</td>
<td>12</td>
<td>A</td>
</tr>
<tr>
<td>Peak forward current</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pulsdauer (Habwertsbreite)</td>
<td>$t_p$</td>
<td>100</td>
<td>ns</td>
</tr>
<tr>
<td>Pulse width (FWHM)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Tastverhältnis</td>
<td>$d.c.$</td>
<td>0.1</td>
<td>%</td>
</tr>
<tr>
<td>Duty cycle</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sperrspannung</td>
<td>$V_R$</td>
<td>3</td>
<td>V</td>
</tr>
<tr>
<td>Reverse voltage</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Betriebstemperatur</td>
<td>$T_{op}$</td>
<td>-40 to +85</td>
<td>°C</td>
</tr>
<tr>
<td>Operating temperature</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lagertemperatur</td>
<td>$T_{stg}$</td>
<td>-40 to +100</td>
<td>°C</td>
</tr>
<tr>
<td>Storage temperature</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Löttemperatur ($t_{max} = 10 \text{s, 2 mm von Gehäuseunterseite}$)</td>
<td>$T_s$</td>
<td>260</td>
<td>°C</td>
</tr>
<tr>
<td>Soldering temperature</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>($t_{max} = 10 \text{s, 2 mm from bottom edge of case}$)</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
**Optische Kennwerte** (Umgebungstemperatur $T_a = 25 \, ^\circ C$)

**Optical Characteristics** (Ambient temperature $T_a = 25 \, ^\circ C$)

<table>
<thead>
<tr>
<th>Parameter Parameter</th>
<th>Symbol</th>
<th>Symbol</th>
<th>Werte Values</th>
<th>Einheit Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Zentrale Emissionswellenlänge$^1$</td>
<td>$\lambda_{\text{peak}}$</td>
<td>840</td>
<td>850</td>
<td>860</td>
</tr>
<tr>
<td>Spektralbreite (Halbwertsbreite)$^1$</td>
<td>$\Delta \lambda$</td>
<td>–</td>
<td>4</td>
<td>–</td>
</tr>
<tr>
<td>Spectral width (FWHM)$^1$</td>
<td>$\nu_{\text{op}}$</td>
<td>7</td>
<td>10</td>
<td>13</td>
</tr>
<tr>
<td>Spitzenausgangsleistung$^1$</td>
<td>$I_{\text{th}}$</td>
<td>0.5</td>
<td>0.75</td>
<td>1.0</td>
</tr>
<tr>
<td>Peak output power$^1$</td>
<td>$V_{op}$</td>
<td>2.0</td>
<td>2.5</td>
<td>3.0</td>
</tr>
<tr>
<td>Schwellstrom Threshold current</td>
<td>$l_r, l_f$</td>
<td>–</td>
<td>1</td>
<td>–</td>
</tr>
<tr>
<td>Betriebsspannung$^1$</td>
<td>$l_r, l_f$</td>
<td>–</td>
<td>1</td>
<td>–</td>
</tr>
<tr>
<td>Operating voltage$^1$</td>
<td>Apertur size</td>
<td>$w \times h$</td>
<td>–</td>
<td>200 × 2</td>
</tr>
<tr>
<td>Strahlendivergenz (Halbwertsbreite) Beam divergence (FWHM)</td>
<td>$\theta_{\parallel} \times \theta_{\perp}$</td>
<td>–</td>
<td>9 × 25</td>
<td>–</td>
</tr>
<tr>
<td>Temperaturkoeffizient der Wellenlänge Temperature coefficient of wavelength</td>
<td>$\partial \lambda / \partial T$</td>
<td>–</td>
<td>0.25</td>
<td>–</td>
</tr>
<tr>
<td>Temperaturkoeffizient der opt. Ausgangsleistung Temperature coefficient of optical power</td>
<td>$\partial P_{\text{opt}} / P_{\text{opt}} \partial T$</td>
<td>–</td>
<td>-0.4</td>
<td>–</td>
</tr>
<tr>
<td>Thermischer Widerstand Thermal resistance</td>
<td>$R_{th, JA}$</td>
<td>–</td>
<td>160</td>
<td>–</td>
</tr>
</tbody>
</table>

$^1$ Standardbetriebsbedingungen beziehen sich auf eine Pulsdauer von 100 ns bei einer Frequenz von 1 kHz und einem Betriebsstrom von 10 A bei $T_a = 25 \, ^\circ C$.

Standard operating conditions refer to pulses of 100 ns pulse width at 1 kHz rate with 10 A operating current at $T_a = 25 \, ^\circ C$. 

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Optical output power $P_{\text{opt}}$ and forward voltage $V_F$ vs. forward current $I_F$ ($T_A = 25 \, ^\circ\text{C}$)

Optical spectrum, relative intensity $I_{\text{rel}}$ vs. wavelength $\lambda$ ($T_A = 25 \, ^\circ\text{C}, P_{\text{op}} = 10 \, \text{W}$)

Far-field distribution parallel to junction $I_{\text{rel}}$ vs. $\theta_||$ ($T_A = 25 \, ^\circ\text{C}, P_{\text{op}} = 10 \, \text{W}$)

Far-field distribution perpendicular to junction $I_{\text{rel}}$ vs. $\theta_\perp$ ($T_A = 25 \, ^\circ\text{C}, P_{\text{op}} = 10 \, \text{W}$)

Opto Semiconductors

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