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Design and Manufacturing III

Automated Exchange Transfusion Device
Final Report

Exchange Transfusion Team, Project 17

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ABSTRACT – At the Komfo Anokye Teaching Hospital in Kumasi, Ghana, exchange transfusions are performed manually in the Mother Baby Unit (MBU). Currently, these transfusions take multiple hours to perform. Because an automated process is absent, a doctor or nurse must be present throughout the entire transfusion to constantly turn valves and administer or remove blood from syringes. A low-cost, easy to operate, automated exchange transfusion device is essential in order to increase the number of available human resources in the MBU, thus providing doctors and nurses with more time to care for other patients.

EXECUTIVE SUMMARY

On average, one neonatal exchange transfusion (ET) is performed at the Komfo Anokye Teaching Hospital (KATH) in Kumasi, Ghana each day. The medical personnel at KATH currently perform neonatal ETs manually, which entails manually rotating valves and operating syringes. The entire ET process takes approximately 90-120 minutes to complete; therefore, occupying much of the doctor's time. In order to increase the number of available medical personnel in the Mother Baby Unit (MBU) at KATH, it is crucial that KATH acquires a low-cost, easy to operate, automated ET device.

The primary focus of this design is to create an automated ET device; however, there are several other customer requirements that must be taken into account during the design process to ensure that the device can be utilized by KATH in Ghana. For the ET device to be feasible, it first needs to be safe and consistent. The ET should also be easy to operate. The device must also have variable blood flow rate settings to be compatible for individual patient needs. Safety measures that are used when dealing with medical procedures, such as using disposable syringes will be taken into account. The ET device must also be inexpensive due to the minimal allotted budget for the MBU.

Various brainstorming sessions took place to generate a viable automated ET device. After evaluating the brainstorming concepts, five of the best concepts were analyzed and further developed. The following five concepts were chosen for further evaluation: the out-of-phase slider, the in-phase slider, the gravity-piston, the compressor system, and the automated four-way stop cock. Once these concepts were developed, a Pugh chart, as well as a thorough analysis was used to determine the alpha design concept—the in-phase slider.

The initial in-phase slider design was elaborated in detail for the prototype, primarily because of its versatility. It allows for an adjustable volume flow rate because of its user interface, it can be used with various syringes, it contains a feedback control system, it requires only one insertion point, and it can easily be transitioned into a manual process if needed. It utilizes a linear actuator as the motor, and a sliding mechanism to support the syringes and their movement. The in-phase slider does have a few disadvantages, however, such as the scarcity of component parts in Ghana (i.e. circuit board and motor).

Through means of measuring and testing various components of the in-phase slider design, specific design parameters were determined, such as the component dimensions. A thorough analysis of the in-phase slider design was then performed to determine the manufacturability and functionality of the design. Each component of the in-phase slider was evaluated in detail, in which the material selection and manufacturing process was validated by means of a detailed parameter analysis.

After a detailed evaluation of the in-phase slider design, the fabrication plan for the prototype was established; therefore, providing a step-by-step plan for producing the design. Using this plan, the prototype was manufactured, and numerous validation tests were performed on it to insure functionality and reliability. Different colored water was used to resemble fresh and waste blood to validate fluid flow functionality, as this was the most important requirement our device had to meet. After all testing was completed, we concluded that our device could successfully perform an exchange transfusion and met all of our key engineering specifications; therefore it was ready to be displayed at the Design Expo.

The in-phase slider functions exceptionally well when considering the time and budget constraints that had to be overcome. However, a few changes that could potentially be made if the process was to be done over again would be to: find a smaller, more inexpensive motor, add a casing to further protect the circuit board and user interface, and use proper medical tubing for more sufficient flow validation. In conclusion, the device can perform an automatic exchange transfusion successfully, and all key customer requirements and engineering specifications have been met.

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An Automated Exchange Transfusion Device for the Komfo Anokye Teaching Hospital in Kumasi, Ghana

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Abstract – **At the Komfo Anokye Teaching Hospital in Kumasi, Ghana, exchange transfusions are performed manually in the Mother Baby Unit (MBU). Currently, these transfusions take multiple hours to perform. Because an automated process is absent, a doctor or nurse must be present throughout the entire transfusion to constantly turn valves and administer or remove blood from syringes. A low-cost, easy to operate, automated exchange transfusion device is essential in order to increase the number of available human resources in the MBU, thus providing doctors and nurses with more time to care for other patients.**

Index Terms – automated technology, exchange transfusion, KATH, Kumasi, medical device, neonate

INTRODUCTION

This paper proves that the consolidation of undergraduate engineering design projects with international service learning provides an invaluable experience for undergraduate students. It requires students to explore cultural diversity to better understand the requirements of their engineering design project. A union between the engineering curriculum and service learning offers the students a skill set that incorporates critical thinking beyond the scope of a typical undergraduate student and cooperative learning. This paper will reveal the benefit of uniting service learning and undergraduate engineering design projects by divulging the project that four undergraduate students at the University of Michigan embarked on.

Background Information

The availability of medical personnel is a great issue in Ghana. In 2004, the World Health Organization (WHO) stated that Ghana only has 2.00 physicians per 10,000 people[i]. WHO also affirmed that there are only 9.00 nursing and midwifery personnel per 10,000 people in Ghana. This reaffirms that there are far more neonates that require medical attention than there are doctors/nurses who can offer medical attention. The Komfo Anokye Teaching Hospital (KATH) in Kumasi, Ghana is a prime example of a facility where this problem is prevalent.

KATH is the lone hospital for over 100,000 people in urban Ghana. On average, one neonatal exchange transfusion (ET) is performed at KATH each day (J. Adabie, personal interview, February 7, 2009). The medical personnel at KATH presently perform neonatal ETs manually, which entails manually rotating valves and operating syringes, which is depicted in Figure 1 below. The entire ET process takes approximately 90-120 minutes to complete, thus occupying much of the doctor's valuable time. The time a doctor spends performing an ET is time he is not spending with other neonates in critical condition. The lack of valuable human resources, primarily doctors, is one of the issues contributing to the high mortality rate in neonates. There is a 2.7% neonatal mortality rate in Ghana. 28.5% of deaths among children under 5 years of age is due to neonatal causes[i]. The limited medical staff prohibits many neonates from receiving the care they need; ETs play a major role in the time limitation problems for the medical personnel at KATH.

FIGURE 1
MANUAL EXCHANGE TRANSFUSION PROCEDURE BEING PERFORMED TO A NEONATE AT KATH

Objectives on Behalf of KATH

As previously mentioned, the limited number of medical personnel in the Mother Baby Unit (MBU) at KATH prevents the patients from receiving the care they need. The shortage of available medical personnel is partially attributed to the time consuming ET process presently implemented at KATH. A low-cost, easy to operate, automated ET device is a sufficient solution in order to increase the number of available medical personnel in the MBU, thus providing doctors and nurses with more time to care for other patients. This will in effect increase the number of patients attended to each day, and hopefully, decrease the mortality rate of neonates in Ghana. As a group of four undergraduate engineering students pursued the automated ET device

design project, the medical personnel at KATH established the following set of specifications for the project:

- Produce an "automated" exchange transfusion device; therefore, the presence of the medical personnel will not be necessary for the entire ET. This will allow the medical personnel to care for other patients.
- Create an inexpensive ET device. Funding is extremely limited in the medical field in Ghana; therefore, a cost of less than 200 USD should be targeted for this application.
- Design the automated ET device to utilize local material resources and medical equipment available in Ghana.
- Provide an easy-to-use device, where the process does not require medical personnel beyond the start and end of the transfusion process. This will offer medical personnel with more time to care for other patients in the MBU. The device must also be easily employed by medical personnel.
- Institute an adjustable volume flow rate for the automated ET device. A suitable volume flow rate of blood flowing into and out of a neonate is 3 mL/min [3]; however, the volume flow rate may need to be increased or decreased depending on the type of blood being infused into the neonate due to the varying fluid viscosities of different blood components.

Objectives to Foster Global Awareness

The Kumasi community will not be the only people affected by this project. The consolidation of such a service learning project and engineering design project will provide the students with an excellent opportunity to foster global awareness and assist people in need. Student learning will occur, not only in the technical sense, but also in more tangible ways. To impose such service learning on the students, the following objectives were established:

- Broaden their problem solving skill set to encompass more than just technical aspects, but to also consider cultural, environmental, and social aspects.
- Increase their social network by interacting with a variety of different individuals and professionals.
- Gain exposure to different disciplines and cultures through their communication with various individuals (i.e. medical doctor, electrical engineer, people of Ghana).
- Enhance their understanding of medical technologies and the medical guidelines associated with designing medical devices.
- Increase their insight on the ethical dilemmas encountered due to cultural and societal differences.
- Improve their ability to adapt to unforeseeable details.
- Grow in their knowledge that people, not technologies, are the ultimate focus of engineering through socially relevant projects.

Relationship of Participants

The students completed their capstone mechanical engineering design course under the instruction of Dr. Kathleen Sienko, a professor at the University of Michigan's mechanical engineering department, who is also an organizer of the University's Global Intercultural Experience for Undergraduates (GIEU). The GIEU provides opportunities for students to go

abroad to develop intercultural skills, global perspective, and cultural sensitivity. Students bring back with them a greater understanding of global situations and processes. One such observation brought back to the University was the aforementioned situation in the Mother Baby Unit (MBU) at the Komfo Anokye Teaching Hospital (KATH) in Kumasi, Ghana. The dire situation at the MBU eventually made its way to the four-student mechanical engineering team who decided to do something by designing the automated exchange transfusion device.

DESIGN DESCRIPTION

The design of the automated exchange transfusion device consists of both mechanical and electrical subsystems. The purpose of the mechanical subsystem is to actuate the syringe plungers and cause the flow of blood to and from the patient. A secondary mechanical subsystem is that of the blood pathway between the syringes, blood bags, and patient. The blood pathway subsystem is a unique configuration of tubing, one-way valves, and Y-junction connectors. The purpose of the electrical subsystem is to power and control the linear actuator and its function of driving the mechanical sliding mechanism. The result of the three subsystems is a mechatronic system that completes the overlying objective of exchanging the blood of a neonatal patient.

System Configuration

The configuration begins with an IV insertion point which consists of a 23-gauge needle into the umbilical vein connected to a 16-outer diameter gauge catheter within the umbilical cord. The catheter leads out of the umbilical cord to a three-way junction which splits the blood flow into two directions. In one direction is a waste-blood suction syringe and waste-blood disposal bag. In the other direction is a fresh-blood suction syringe and fresh-blood bag. Both waste-blood and fresh-blood sides of the system contain three-way junctions between the respective bag of blood, the respective suction syringe, and the initial three-way junction just outside of the umbilical cord. All junctions in the system are open to the flow in both directions. Equipped on two of the three sides of each junction are disposable one-way valves configured in a way to allow functioning of the system and to eliminate the need for manual toggling of the three-way junctions.

Process Description

The ET process is prefaced by priming the entire system with the blood intended to be transferred. The tubing, junctions, and valves need to be filled with blood before the process begins. The suction syringes also begin in their fully compressed position and operate simultaneously and in-phase with each other. Both syringe heads are connected to a slider mechanism which operates in the same direction as the syringe plunger and is powered by a linear actuator. Attaching the syringe heads to the same slider allows the syringes to operate simultaneously and in-phase with each other.

The cycle begins when the slider, and thus the syringe plungers, move back and the syringes intake fluid into their bodies. On the side with the fresh blood bag, fresh-blood is removed from the bag past the one-way valve and into the syringe. During this phase, blood is not removed from the neonate with this syringe because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood is removed from the neonate past the initial three-way junction, and into the syringe. During this phase blood is not removed from the waste-blood

disposal bag because of the one-way valve configuration. This phase continues until the syringe plungers are fully extended back and the syringes are full with their respective fresh-blood and waste-blood.

The next phase of the cycle begins when the slider, and thus the syringe plungers move forward and the syringes force fluid out of their bodies. On the side with the fresh-blood bag, fresh-blood in the syringe is forced out past both three way junctions and into the neonate. During this phase fresh-blood is not forced back into the fresh-blood bag because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood in the syringe is forced out into the waste-blood disposal bag. During this phase waste-blood is not forced back into the neonate because of the one way valve configuration. This phase continues until the syringe plungers are compressed all the way and the syringes are empty of all their respective fluids. When this phase completes, the first phase begins again, and the cycle repeats. A complete cycle of the process is depicted in the diagram of the full system shown in Figure 2 below.
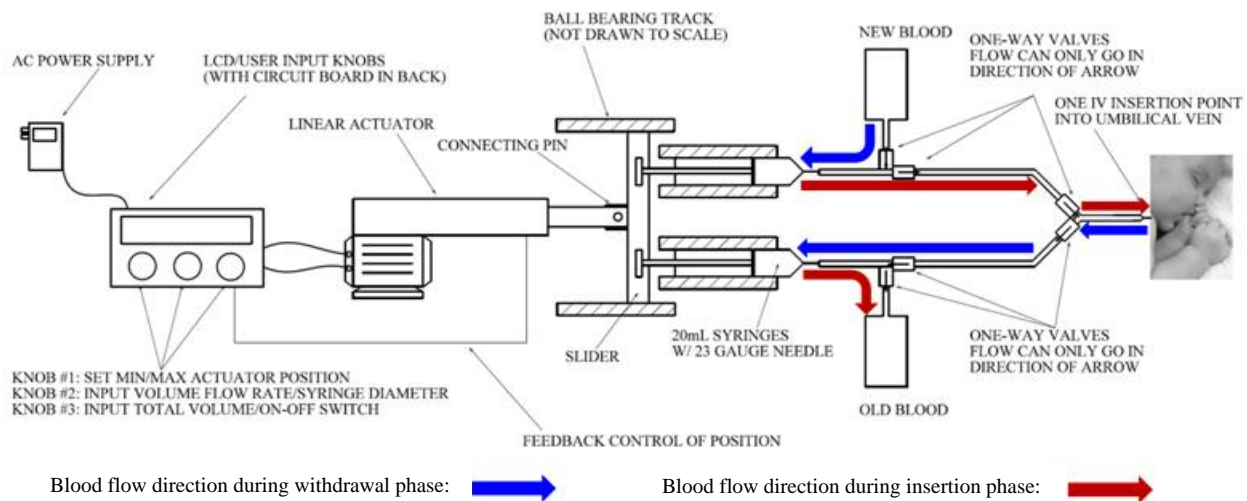
FIGURE 2

DIAGRAM OF PROTOTYPED AUTOMATED EXCHANGE TRANSFUSION SYSTEM.

Mechanical Subsystem

The mechanical subsystem consists of a holding block, a slider, a track, and a linear actuator. The subsystem can be seen in Figure 3 with two 25 mL syringes in place. Not included in the picture is the linear actuator, which is also a part of the mechanical subsystem, and is connected to the slider with a pin. The mechanical subsystem of the final prototype is shown in Figure 4 as well.
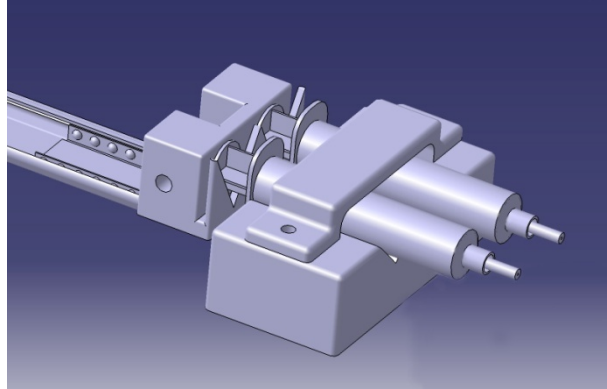
FIGURE 3

EARLY CAD DESIGN OF THE MECHANICAL SUBSYSTEM CONTAINING
THE SYRINGE HOLDING BLOCK AND BRACKET, SLIDER, AND TRACK.
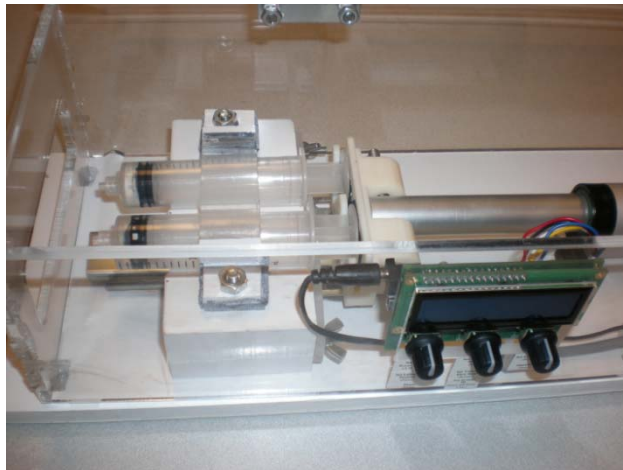
FIGURE 4

PROTOTYPE OF AUTOMATED EXCHANGE TRANSFUSION DEVICE.

The linear actuator that was used to actuate the slider was purchased and was the most inexpensive on the market. It is depicted in Figure 5. The linear actuator is connected to the slider by a pin and the velocity of the shaft will be micro-controlled to obtain the desired volume flow rate of the syringe, which will depend on the diameter of the syringe. The motor itself is 12 V DC and at the loads expected for our application the shaft can actuate up to 0.59 in/sec. For the maximum desired flow rate and for a 25 mL syringe diameter, a linear velocity of 0.00088 in/sec is needed. This posed a lower limit, rather than an upper limit problem with the motor speed. However, after analysis of the voltage resolution out of the micro-driver, we calculated that the motor would be able to handle speed low enough for our application.

FIGURE 5
LINEAR ACTUATOR USED IN THE EXCHANGE TRANSFUSION DEVICE PROTOTYPE.

The linear actuator comes equipped with a position potentiometer as well, which comes into play with the feedback control schedule, which is described in the *Electrical/Circuit Subsystem* section below.

Electrical/Circuit Subsystem

While the mechanical subsystem is the bones of the exchange transfusion system, the electrical/circuit subsystem is the lifeblood. An electrical aspect to an automated exchange transfusion device is absolutely essential. If the device is to be automatically controlled and powered; a circuit, power source, and controller must be implemented. The overlying specification that must be met in order for the ET device to insure the safety of the neonate is the control of the blood volume flow rate. The component of the system that makes this happen is the microprocessor. The other electrical components exist to provide support, or input to the microprocessor (except the buzzer alarm which serves as its own engineering specification).

The automated ET device was designed to accommodate the various needs of the neonate and the medical personnel. To oblige to the needs of the medical personnel, the automated ET device must also be compatible with an assortment of syringes ranging from 20 to 25 mL in volume. The ET device will also be required to stop when a specified volume of blood has been transfused. In order to adapt to all of these variables, the automated ET device must have a user input to inform the ET device of the necessary parameters. From the user input, the ET device's operation will adjust accordingly.

The user interface consists of three rotary encoders and a LCD. The three rotary encoders will allow the medical personnel to input the necessary system parameters. The three encoders and the LCD will be soldered onto the top of the circuit board with the LCD screen just above the three rotary encoders as depicted in Figure 6. The encoders can either be spun or pressed by the user to allow multiple inputs for the three knobs.
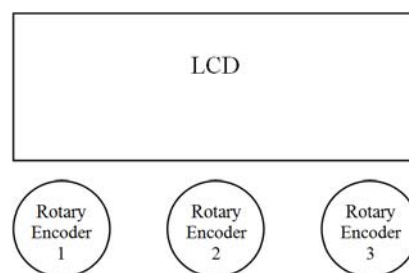

FIGURE 6
DIAGRAM OF THE USER INTERFACE WHICH IS LOCATED ON TOP OF THE ELECTRICAL CIRCUIT BOARD.

Control of the blood volume flow rate will be conducted with a position schedule. When the user inputs a desired blood volume flow rate and syringe diameter into the microprocessor, it calculates the necessary linear velocity of the actuator shaft. With the minimum and maximum positions of the actuator set prior to start, the microprocessor determines where and when exactly the shaft should be between the minimum and maximum set points to achieve the desired blood volume flow rate. This is known as a position schedule.

The linear actuator is equipped with a built-in potentiometer that detects position of the shaft, which is fed back to the microprocessor. If the shaft is detected as being at a position that is ahead of schedule, the microprocessor pauses the motor and the shaft movement until the schedule catches up to the shaft position. If the shaft is detected as being at a position that is behind schedule, instead of speeding the motor up, the microprocessor just shifts the schedule back to coincide with the current schedule.

Tubing Subsystem

The tubing subsystem is the connection between the blood reservoirs, the syringes, and the neonate. It is the pathway for the blood. The pathway that the blood takes from fresh blood bag, to syringe, to neonate; and waste blood from neonate, to syringe, to waste bag, is thoroughly described in the preceding section *Process Description*. While the tubing and its related components such as the one-way valves is considered its own subsystem, its design has already fully been described in the *Process Description* section.

VALIDATION

Once full system assembly was completed, testing began and several different experiments were conducted to verify that the device met all key customer requirements and engineering specifications. To have a better understanding of fluid flow throughout the system, red and blue colored water was used for all system tests to act as the blood coming out of the neonate, and the blood going into the neonate, respectively. Three IV bags were also used for testing purposes: one bag was filled completely with red water to act as the fresh blood bag, one bag was left empty to act as the waste blood bag, and the last bag was filled completely with blue water to act as the neonate. Descriptions of the validation tests and results are described below.

Setup Time and Ease of Use

Before conducting validation tests on the device while it was transfusing fluid, the time required to prepare the system was tested. This preparation time consisted of securing the two syringes to the device, inputting sample desired values for the volume flow rate, total volume to be transfused, and inner diameter of the syringes into the user interface, and prepping the tubing with water to act as blood to eliminate any air bubbles from the tubing before attempting to run the system. To test the preparation time, six random participants were gathered. Each participant took two turns trying to set up the system. For the first attempt, each of the six participants was given personal help by one of our group members to set up the system. On average, the setup time for this test took approximately 16 minutes and 16 seconds. For the second attempt, each participant had to set up the system using only the instruction manual; no help from any of the group members was given. On average, the setup time for this test took approximately 19 minutes and 17 seconds. Both tests resulted in an average setup time of less than the desired 20

minutes, thus ensuring that the setup time and ease of use engineering specifications were both satisfied.

Volume Flow Rate

Once the preliminary tests were completed, it was time to power the system to begin testing the functionality of the device as it was transfusing fluid. Because an adjustable and consistent volume flow rate is so important to the proper functionality of our system, several tests were conducted to ensure that the rate stayed consistent with the input value, and that the system could run properly at volume flow rates between 0 and 10 mL/min. For tests running at 2,4,6,8 and 10 mL/min, the time required to displace 15 mL of fluid was recorded. After calculating the measured volume flow rate, this rate was compared to the desired volume flow rate. For each of the 5 different flow rates tested, an accuracy of over 99.5% was achieved. A table displaying the full results of the experiment can be seen in Table 1. This easily satisfied the adjustable volume flow rate engineering specification. It should be noted that it is reasonable to assume that the system had a constant flow rate at all times because the linear actuator's potentiometer is constantly detecting position to ensure a steady volume flow rate.

TABLE 1
VALIDATION RESULTS OF VOLUME FLOW RATE TEST

| Desired Flow Rate (mL/min) | Vol / Stroke (mL) | Measured | | % Accuracy |
		Time/stroke (min sec)	Flow Rate (mL/min)	
10	15	1 min 31 sec	9.89	98.90%
8	15	1 min 53 sec	7.96	99.56%
6	15	2 min 30 sec	6.00	100.00%
4	15	3 min 46 sec	3.98	99.56%
2	15	7 min 32 sec	1.99	99.56%

Total Volume Transfused

To validate that the total volume to be transfused was correct, the system was run twice for a full two hour cycle at a volume flow rate of 10 mL/min. For these two tests, the fresh blood IV bag was filled with 1800 ± 25 mL of red water. The waste blood IV bag was then filled with only about 50 ± 25 mL of blue water, and finally the IV bag representing the neonate was filled with 1500 ± 50 mL of blue water. After both two-hour tests, the new total volumes from the bags were recorded. For both tests, the fresh blood bag was reduced by 600 ± 50 mL of water, the waste blood bag was increased by 600 ± 50 mL of water, and the bag representing the neonate remained at a constant volume. A picture clearly depicting these results can be seen in Figure 7. From these results, it was concluded that the total volume to be transfused was correct for each system test. For further verification that fluid flow was correct going into and out of the bag representing the neonate, a color change was witnessed near the end of the bag where the fluid was flowing in. The color of the water changed from a solid blue to a more purple color as the "fresh blood" red water was injected and mixed with the old, "waste blood" blue water. This color change proved that "fresh blood" was correctly entering the neonate throughout the transfusion.

USER INPUTTED VOLUME TO BE
TRANSFUSED = 600mL

WASTE BLOOD BAG   FRESH BLOOD BAG

$V_{initial} = 50mL \pm 25mL$ $V_{initial} = 1800mL \pm 25mL$
$V_{final} = 650mL \pm 25mL$ $V_{final} = 1200mL \pm 25mL$

$\triangle V = 600mL \pm 50mL$ $\triangle V = 600mL \pm 50mL$

"BABY" BLOOD BAG

$V_{initial} = 1500mL \pm 50mL$
$V_{final} = 1500mL \pm 50mL$

$\triangle V = 0mL \pm 100mL$
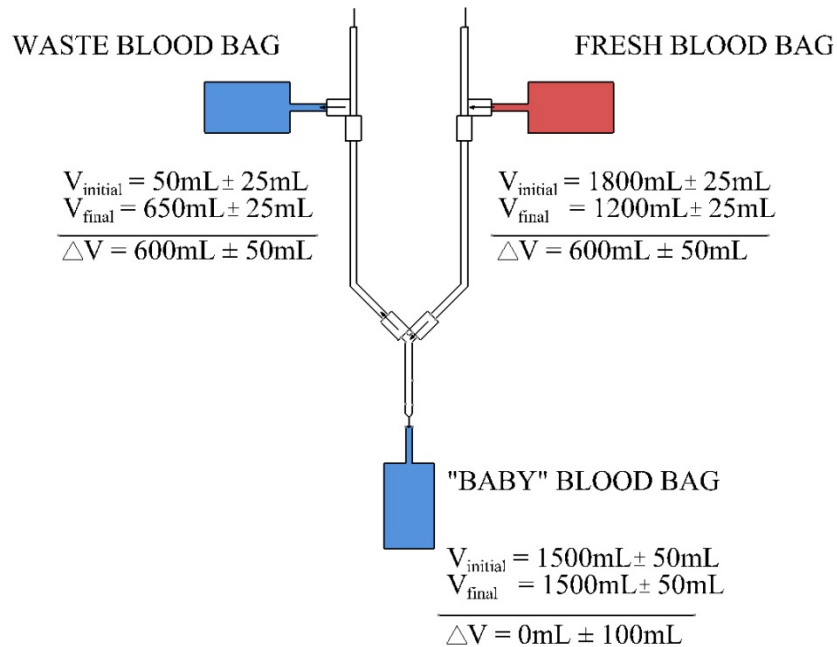
FIGURE 7
VALIDATION RESULTS OF BLOOD DISPLACEMENT TEST

In summary, the automated exchange transfusion device successfully functions as intended. The one way valves allow the fluid to flow properly into and out of the IV bags and the neonate, and the motor and pumping mechanism allow the transfusion to take place at various, but constant volume flow rates. The desired total volume to be transfused is accurate, and the device is compatible with multiple size syringes. Also, with proper preparation, air bubbles will be eliminated from the system thus eliminating any kind of danger regarding air embolism. All of these factors combined allow us to conclude that the automated exchange transfusion device functions safely and successfully.

PROTOTYPE COST

It was also necessary to design the automated exchange transfusion device to be made up of low-cost, locally available materials to Ghana. KATH is a relatively low-budget hospital in urban Ghana and the utilization of low-cost, locally available materials would greatly increase the purchasability of the device. Initially, a target of 200 USD or less per unit was set. This cost would include all components needed to complete the transfusion except basic medical supplies such as syringes, connecting tubing, and IV blood bags. The cost of the prototype rose slightly over the target, however the cost per unit after developing the prototype and with diminishing costs that occur when mass manufacturing, the device is expected to be less than 120 USD.

CONCLUSIONS AND REFLECTIONS

In summary, all of the objectives on behalf of KATH have been met. A successful automated transfusion device has been fabricated, and all key customer requirements were fulfilled. The device functions as intended, and the important specifications such as an adjustable volume flow rate, a simple and quick setup time, and an automatic shut off function were all accomplished. The objectives to foster global awareness have also been met successfully. The four University of Michigan undergraduate students who worked on the project united service learning with an engineering curriculum to develop a medical device using technical experience, while also gaining knowledge and exposure to different disciplines and cultures throughout the process.

To create a medical device that would potentially be used in a foreign country without ever actually visiting that country, critical thinking beyond the scope of a typical undergraduate student and cooperative learning was essential. The students involved with the project had to broaden their problem solving skill set to encompass more than just technical aspects. Cultural, environmental, and social aspects in Ghana all had to be considered as well. Conversations with medical staff at the KATH hospital in Ghana helped give valuable insight to these non-technical aspects of the project. Funding for medical equipment and safety guidelines are different in Ghana than they are in the United States, so exploration and understanding of those differences helped the group to develop a more practical exchange transfusion device for Ghanaian life.

The engineering students involved with the project also had to look beyond the technical aspects of the project to increase their insight on the ethical dilemmas encountered due to cultural and societal differences. Funding in Ghana for medical equipment is limited, so decisions had to be made regarding when to sacrifice cost limitations for safety and reliability. Because the device would potentially have human lives in its hands if it was ever implemented, there were constant ethical considerations that had to be addressed throughout the design process. In the end, safety with a greater cost was ruled to be more important than sacrificing safety to lower cost. This was most importantly shown in the selection of the motor, where a more expensive linear actuator was chosen over a cheaper, but far less consistent gear/crank mechanism.

The ethical issue of this medical device being automatic was also addressed. Because medical staff availability is so limited in Ghana, the staff could have a tendency to take the device for granted if it were to be implemented and the neonate receiving the transfusion could be left alone for too long. Although the device will be completely automatic beyond the start-up and prepping, the vitals of the neonate along with proper functionality will still need to be frequently monitored to ensure safety of the neonate at all times.

For the automated exchange transfusion device to be successfully implemented as a real medical device in the future at the KATH hospital in Ghana, further work would absolutely be essential. The device was developed with available resources to the students working on it, and it was created under a very limited budget. To become a medical device to be used on neonates in Ghana, a greater material selection process would need to be conducted to ensure the safety and the durability of the device to the highest extent. Animal testing with the device would also need to be done before it could ever be implemented in a hospital setting.

APPENDICES

Appendix A: Updated Design Review III Document

Nomenclature

Important symbols and acronyms used throughout this report are presented in Table 1.

Table 1. Important Nomenclature

Symbol	Unit	Definition
AS	---	Additive Solution
CV	---	Central Vein
ELBW	---	Extremely Low Birth Weight
FDA	---	Food and Drug Administration
FFP	---	Fresh Frozen Plasma
HBV	---	Hepatitis B Virus
HCV	---	Hepatitis C Virus
HIV	---	Human Immunodeficiency Virus
KATH	---	Komfo Anokye Teaching Hospital
MBU	---	Mother Baby Unit
PRBC	---	Packed Red Blood Cell
RBC	---	Red Blood Cell
VLBW	---	Very Low Birth Weight
WB	---	White Blood
WBC	---	White Blood Cell
WHO	---	World Health Organization

Introduction

The following sections will discuss the problem the Mother Baby Unit (MBU) of the Komfo Anokye Teaching Hospital (KATH) in Kumasi, Ghana is currently facing.

Problem Definition

The availability of medical personnel is a great issue in Ghana. In 2004, the World Health Organization (WHO) stated that Ghana only has 2.00 physicians per 10,000 people [1]. WHO also affirmed that there are only 9.00 nursing and midwifery personnel per 10,000 people in Ghana. This reaffirms that there are far more neonates that require medical attention than there are doctors/nurses who can offer medical attention. KATH is a prime example of a facility where this problem is prevalent.

Dr. John Adabie Appiah, the Child Health Directorate at KATH, is the mentor for this project, and he is one of the doctors in charge of the Mother Baby Unit (MBU) at KATH. KATH is the lone hospital for over 100,000 people in urban Ghana. On average, one neonatal exchange transfusion (ET) is performed at KATH each day (J. Adabie, personal interview, February 7, 2009). The medical personnel at KATH presently perform neonatal ETs manually, which entails manually rotating valves and operating syringes. The entire ET process takes approximately 90-120 minutes to complete, thus occupying much of the doctor's valuable time. The time a doctor spends performing an ET is time he is not spending with other neonates in critical condition. The lack of valuable human resources, primarily doctors, is one of the issues contributing to the high mortality rate in neonates. There is a 2.7% neonatal mortality rate in Ghana. 28.5% of deaths among children under 5 years of age is due to neonatal causes [1]. The limited medical staff prohibits many neonates from receiving the care they need; ETs play a major role in the time limitation problems for the medical personnel at KATH.

Current Process

Presently, ETs are performed manually at KATH. The following provides the step-by-step procedure currently used for neonatal ETs at KATH: (1) Attach the apparatus to the neonate's umbilical vein with the valves oriented, so that blood can be extracted directly from the neonate to the syringe. (2) After the syringe is filled, toggle the valves to close the path between the neonate and the syringe and to open a path between the syringe and the disposal container. (3) Completely empty the extracted blood from the syringe into the disposal container. (4) Toggle the valves to close the path to the disposal container and open a path between the syringe and the fresh blood container. (5) Fill the syringe with the fresh blood. (6) Toggle the valves to close the path between the syringe and the containers of blood and to open the path between the syringe and the neonate. (7) Transfer the fresh blood from the syringe into the neonate. (8) Repeat the process until a sufficient amount of blood has been transferred into the neonate. A visual depiction of the process explained above can be seen in Figure 1 below.
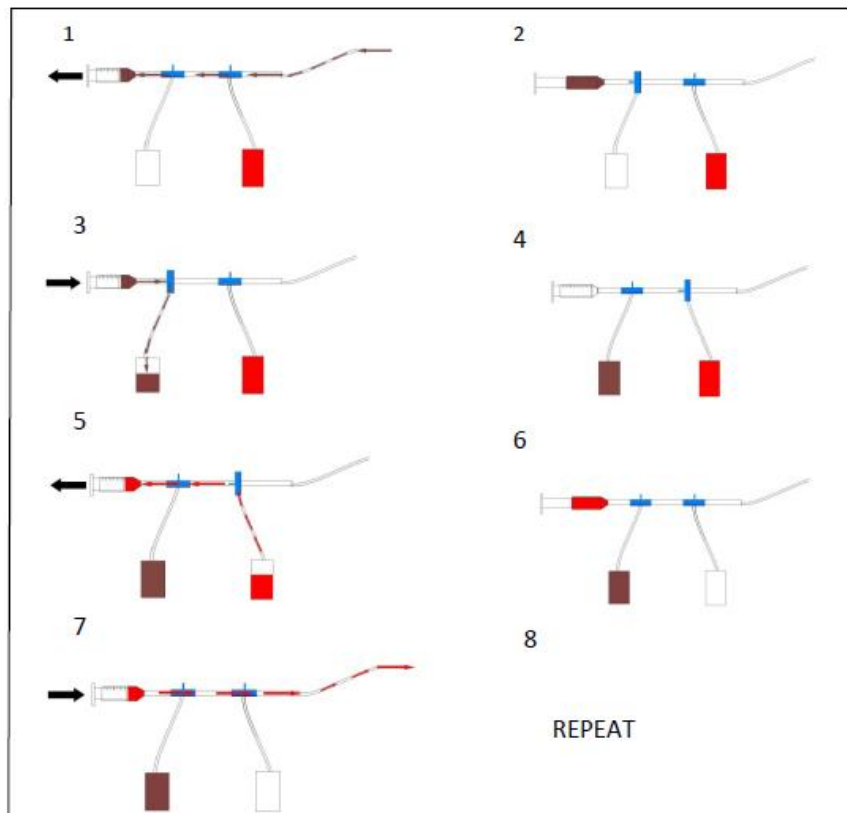
Figure 1. Current blood transfusion process (procedure described above).

An infant must be monitored by a doctor or nurse at all times during this process, which may take several hours. It would be very beneficial for KATH to have an automated ET device; therefore, the presence of a doctor and/or nurse is no longer necessary. An automated ET device would provide the doctor/nurse more time to care for other patients.

As a result of the problems stated above, we have been asked to develop a low-cost, easy to operate and maintain, automated ET device for KATH.

Background

Blood transfusion is the process of transferring blood or blood-based products from one individual into the circulatory system of another. In the year 2000, there were over 90 million blood transfusions worldwide [2]. The neonatal population receives the majority of all blood transfusions. There are no strict guidelines outlining how to perform a blood transfusion procedure. In this section, we will investigate the need for blood transfusions, the requirements associated with blood transfusions, specifically in neonates, the equipment used to perform a blood transfusion, and the transfusion methods in developing countries.

Individual Blood Components for Transfusion

Whole blood is made up of the following components: red blood cells (RBC), plasma, and platelets. The separation of whole blood into its constituent components is widely practiced for use when only these specific components are required. Each blood component can be processed using different methods and derivatives, thus leading to over ten types of blood products. Neonates are primarily transfused with the following types of blood products: whole blood, packed red blood cells, platelets, and fresh frozen plasma.

Whole blood: Whole blood (WB) is desegregated blood collected into an approved container containing an anticoagulant-preservative solution [3]. WB contains RBCs, plasma, clotting factors, platelets, and approximately 10^9 white blood cells (WBC) [2]. Table 2 provides the composition of WB, as well as the hematocrit level of WB (p. 4). WB is transfused to increase the oxygen-carrying capacity of the patient's blood to maintain adequate tissue oxygenation by replenishing blood volume and red cell mass [2]. WB provides oxygen-carrying capacity, blood volume expansion, and stable coagulation factors [3]. WB may also be useful for patients with both RBC volume deficits, such as patients with hemorrhages. The use of WB is discouraged in most situations, as the use of specific blood components is more appropriately tailored to the unique needs of the patient [2]. WB should not be used if a patient is at risk of circulatory overload because whole blood has a higher volume than red cells [3].

Packed red blood cells: Packed red blood cells (PRBCs) are the most commonly transfused blood component [2]. Table 2 provides the composition of PRBCs, as well as the hematocrit level of PRBCs (p. 4). PRBCs have the same red cell volume as WB; therefore, PRBCs have the same oxygen-carrying capacity as WB in a significantly reduced volume. PRCBs are mainly used for resolution of symptomatic anemia and improvement of tissue oxygenation [4]. PRBCs require more time for transfusion compared to WB, because PRBCs have a high ratio of red cells to plasma, which increases viscosity [3]. Fresh RBCs, less than 7 days old, are typically used for an infant's initial small-volume transfusion [5]. RBC units with an additive solution (AS) are used for pediatric infusions; however, AS-RBCs are not used routinely in newborns because of concerns about reactive hyperinsulinemia, renal toxicity, and dieresis-associated fluctuations in cerebral blood flow, which could be triggered by the higher concentrations of dextrose, adenine, and mannitol in AS units.

Platelets: Platelets play a significant role in hemostasis by repairing breaks in small blood vessel walls through the formation of blood clots [2]. Deficiencies in the platelet amount and/or function can have unpredictable effects that range from clinically insignificant prolongation of the patient's bleeding time to major life-threatening hemorrhaging. Platelets are typically transfused when a patient has thrombocytopenia or when a patient is experiencing platelet function defects (often caused by decreased platelet production or increased platelet destruction) [2, 3]. Platelets should be administered when the patient's platelet count falls below 50×10^9/L and the patient shows clinical signs of microvascular bleeding. Platelet transfusions should be considered in cases where the platelet count falls below 20×10^9/L, even if there is no clinical evidence of bleeding, because there is a danger of hidden bleeding, such as into the brain tissue [3]. Platelets used for transfusion may come from two different sources: (1) those prepared by separating rich plasma from a routine whole blood donor ("platelet concentrates" or whole

blood derived platelets), and (2) those collected from a single donor using the cytapheresis process ("single donor platelets" or apheresis derived platelets) [2]. The recipient of platelet concentrate is exposed to 5 to 8 times more blood donors per transfusion than a single donor platelet recipient. The composition of platelets is revealed in Table 2 below.

Fresh Frozen Plasma: Plasma is the aqueous, acellular portion of WB, which consists of proteins, colloids, nutrients, crystalloids, hormones, and vitamins [2]. Fresh frozen plasma (FFP) is plasma that is frozen within 6 to 8 hours of collection and stored at -18˚C or colder. The composition of FFP is shown in Table 2 below. Transfusion of FFP is appropriate for the following: (1) replacement of specific coagulation factors not available in concentrate form, (2) management of multiple factor deficiency, which is usually the result of liver disease, and (3) treatment of antithrombin II (ATIII) deficiency [2, 6]. FFP should not be used as a general volume expander because it carries the risk of disease transmission and safer, less expensive products are readily available [2].

Table 2. Individual blood components for transfusion [2].

Component	Approximate Volume (mL)	Composition	Neonatal/Pediatric Dosage	Hematocrit
Whole Blood (WB)	500	250 mL red cells 250 mL plasma 63 mL anticoagulant	10 mL/kg body weight transfused over 2-4 hours	<35% - 40%
Packed red blood cells (PRBCs)	250	CPD or CPDA 200 mL red cells 50 mL plasma	10 mL/kg body weight transfused over 2-4 hours	<50% - 80%
PRBCs (additive solution)	350	200 mL red cells 50 mL plasma 100 mL adenine saline solution	10 mL/kg body weight transfused over 2-4 hours	<50% - 60%
Platelets, apheresis (Single donor)	300	$\geq 3 \times 10^{11}$ platelets $< 10^4\text{-}10^6$ WBCs and plasma	Can be dosed at 10 mL/kg body weight, but most times is dosed by ¼, ½, and whole pheresis units.	---
Platelet concentrate (Random donor)	50	$\geq 5.5 \times 10^{10}$ platelets Variable numbers RBC, WBCs, and plasma	10 mL/kg body weight transfused by gravity, pump, or IV push	---
FFP	180-300	Plasma proteins Immunoglobulins Complement Coagulation factors Albumin	10-15 mL/kg body weight transfused over 1 hour or IV push	---

Neonatal Blood Transfusion

Neonates are among the most frequent to receive blood transfusions [7]. Preterm neonates comprise the most heavily transfused group of patients [4]. Approximately 85% of extremely low birth weight (ELBW), less than 1.0 kg newborns receive a transfusion within the first 2 weeks of life [4, 8]. As many as 80% of all very low birth weight (VLBW), less than 1.5 kg newborns receive multiple RBC

transfusions [7]. An ET in the neonate constitutes a massive transfusion since it involves the replacement of one to two whole blood volumes [2].

Reasons for a blood transfusion in a neonate: The following factors contribute to the decision of whether or not an ET should be performed: gestational age, evidence of hemolysis, degree of anemia, rate of rise of bilirubin, and concurrent clinical conditions [2]. The primary reasons for a blood transfusion in a neonate are hemolysis, volume replacement, respiratory arrest, cardiac disease, and anemia.

Hemolysis: Hemolysis is the destruction of red blood cells, which leads to the release of hemoglobin from within the RBC into the blood plasma [3]. Hemolysis is the underlying cause of kernicerterus, neurological complications caused by rapidly-rising unconjugated bilirubin concentration. The release of hemoglobin into the plasma causes an increase in unconjugated bilirubin concentrations because the immature liver cannot metabolize the breakdown of hemoglobin. ETs are preformed to neonates primarily to prevent kernicerterus. In many countries, the main cause of hemolytic disease of the newborn is due to the Rh D incompatibility between mother and fetus. Hemolytic disease is the most common and clinically important cause of jaundice, a yellowish discoloration of the skin.

Volume Replacement: Acute blood loss is primarily due to the event of an acute hemorrhage or due to phlebotomy loss [2]. A blood transfusion is of the highest priority when dealing with an acute hemorrhage in order to replace the intravascular volume and to stop the bleeding. It is more important to maintain intravascular volume during acute blood loss than to increase the oxygen-carrying capacity of a neonate with hemoglobin levels of 3 to 4 g/dL with RBC transfusions. Signs of moderate-to-severe acute blood loss include the following: irritability or stupor, pallor or mottling, tachycardia, tachypnea, decreased pulmonary intensity, delayed capillary refill, cool extremities, hypotension, metabolic acidosis, and oliguria or anuria.

Phlebotomy is act of obtaining blood from a vein. Weekly phlebotomy loss among preterm infants during the first two weeks of life averages 10% to 30% of total blood volume (10-25 mL/kg) [8]. The replacement of blood drawn for laboratory testing is a critical factor responsible for multiple RBC transfusions in critically ill neonates [2].

Respiratory distress and cardiac disease: In neonates with severe respiratory distress, such as those requiring high volumes of oxygen with ventilator support, it is customary to maintain the hemoglobin level greater than 13.0 g/dL [2]. It is also advised that neonates with severe cardiac disease maintain a hemoglobin level greater than 13.0 g/dL. By maintaining a hemoglobin level greater than 13.0 g/dL, it is believed that the transfused donor RBCs containing adult hemoglobin will provide optimal oxygen delivery through the period of diminished pulmonary function.

Anemia: The decision to transfuse RBCs should be based on hemoglobin and hematocrit levels because all neonates experience a decline in circulating RBC volume beginning during the first weeks of life [2]. Factors other than hemoglobin concentration that must be considered in the decision to transfuse in the chronic anemia setting include: the patient's symptoms, signs, and functional capacities; the presence or absence of cardiorespiratory and central nervous system disease; the cause and anticipated course of the underlying anemia; and alternative therapies, such as iron and/or recominany human erythropoietin therapy. Neonates with late anemia may be considered for a blood transfusion if anemia is thought to be the cause of: poor weight gain, fatigue while feeding, tachypnoea and tacahycardia, or other signs of decompensation [3].

Guidelines for neonatal blood transfusions: Guidelines for transfusing neonates and infants are controversial, and practices vary widely among institutions [2]. Generally, RBC transfusions are given to maintain a level of hemoglobin or hemocrit believed to be most desirable for the existing clinical

conditions. RBC transfusions are given based on the patient's hemoglobin levels, hematocrit levels, and clinical condition. Table 3 provides the basic RBC transfusion guidelines for neonates and infants (p. provided in Table 2 (p. 4).

Table 3. RBC transfusion guidelines for neonates and infants [2, 5].

Hemoglobin Level	Hematocrit Level	Patient's Clinical Condition
Less than 13 g/dL	Less than 40%	• * Severe cardiopulmonary disease and/or • * Severe respiratory distress
Less than 10 g/dL	Less than 30%	• * Moderate cardiopulmonary disease • * Mid-moderate respiratory distress • Perioperative and/or • Critical care
Less than 8 g/dL	Less than 24%	• * Symptomatic anemia • Unexplained breathing disorders and/or • Unexplained poor growth

* It is important that terms used to describe clinical conditions such as "severe" and "symptomatic" be defined to fit local practices and needs for each infant being considered for transfusion.

Infusion rates: Infusion rates vary based on the clinical condition of the patient, as well as the type of blood component being infused.

RBC infusion rate: Acceptable RBC replacement rates in neonates are typically 5 to 10 mL/hr/kg [5]. Neonates should not receive a blood transfusion that lasts over 4 hours; therefore, transfusions of RBCs in volumes of 10 to 20 mL/kg over 2 to 4 hours are generally tolerated for neonates and infants [2]. With rapid blood loss or in massive transfusion situations, rates as fast as 1.5 mL/min/kg can be used; however, the patients should be carefully monitored for drops in ionized calcium levels and blood pressure [5]. RBC transfusions over 25 mL/kg, or greater than one blood volume in 24 hours, require careful attention to possible metabolic and thermal alterations [2]. The rate of infusion should not exceed 2 mL/kg/hour in the presence of cardiac failure [4]. A transfusion of 10 mL/kg of AS-RBC would be expected to raise the newborn's hematocrit level by 7-8%; however, a transfusion of 10 mL/kg of CPDA-1 PRBCs would be expected to raise the newborn's hematocrit level by 9-10%.

Other blood components infusion rate: The recommended dose of platelets for neonates is 1 unit of platelets per 10 kg of body weight, which amounts to 5 mL/kg [4]. The infusion of 5 mL/kg of platelet concentrate should be able to raise the blood platelet count by at least 50×10^9/L in a neonate [9]. A neonate may receive platelets in volumes of 10-20 mL/kg in cases of severe thrombocytopenia [4]. The volume of FFP to be transfused to a neonate is typically between 1 and 20 mL/kg.

Risks associated with neonatal blood transfusion: ETs have their medical risks, which must be acknowledged during the design process of any blood transfusion device. The risks involved are generally more consequential for neonates than for adolescents or adults, since their bodies are less developed and more fragile. Appendix A-1 provides a list of various complications that arise from ETs in neonates. Some of the complications listed in Appendix A-1 are explained in detail below.

Infections: Although it is mandatory to test collected blood for HIV, HBV, and HCV, syphilis, and malaria, transfusion-transmitted infections are still a considerable risk due to the insensitivity of the screening [2].

Air embolisms: Air embolisms are another serious concern to design against [10]. An air embolism is a medical condition caused when gas bubbles enter the blood stream and can cause hypoxia, stroke, or heart attack. This is usually caused externally and by the misuse of the syringe. Therefore, a procedure will be generated to empty the functioning syringe(s) of any and all air bubbles before the automatic blood transfusion procedure begins. This is standard practice whenever a syringe is used for any purpose by a professional.

Thromboembolism: Another cardiovascular risk involved with ETs is the occurrence of thromboembolism [2]. Similar to an air embolism, thromoboembolism occurs when a blood clot inside a blood vessel breaks off due to fluid flow forces and enters into the circulatory system and obstructs blood flow.

Volume Overload: Neonates are at increased risk of fluid overload from transfusion because the volume of the blood component issued may exceed the volume that a neonate can safely endure [4].

Mechanics of manual exchange transfusion in neonates: There are two general techniques for ET: the isovolumetric method and the discontinuous technique.

Isovolumetric method: The isovolumetric method, first described by Wallerstein, entails simultaneously withdrawing and replacing blood in order to avoid sudden changes in arterial pressure [2]. Due to the simultaneous withdrawal and replacement of blood, there must be two sites of vascular access. The umbilical vein is typically used to withdraw blood from the neonate; however, if the umbilical vein cannot be cannulated, peripheral blood vessels may be used to perform the blood transfusion. The isovolumetric method carries the risk of a second catheter; however, it eliminates the risk of swings in blood volume and pressure.

Discontinuous technique: The discontinuous technique described by Diamond et al. is the most commonly ET method [2]. Small aliquots of blood are withdrawn and replaced through a single catheter with a special four-way stopcock. Typically, no more than 5% of the neonate's blood volume (5 mL/kg body weight) is removed or replaced during a 3 to 5-minute cycle. The total time duration for a double-volume exchange is 90 to 120 minutes or 45 to 60 minutes for a single-volume exchange. This process is much like the one used in Ghana.

Technical considerations for neonatal blood transfusion: There are many technical requirements that must be met by the mechanical devices used to transfuse blood. Neonates have small veins; therefore, the needle size and catheter size must be considered in order to minimize hemolysis. The temperature of the blood is also an impacting factor, which can be easily affected by the transfusion device.

Needle: Needles come in a variety of sizes and designs. Stainless steel needles are primarily used for neonatal blood transfusions, because they are easy and less painful to insert [5]. 20 to 25 gauge needles should be used to withdraw and replace blood from a neonate, because such small gauge needles minimize the likelihood of producing hemolysis in the neonate [2].

Catheter: The catheter should be large enough to achieve good flow rates but not so large that it causes mechanical damage to the vein [5]. Short, long diameter catheters were confirmed to have high flow rates with the flow under a pressure 17 times greater than in a longer, small diameter catheter [2]. There are various types of catheters; however, over-the-needle catheters causes less trauma to the vein for a short-term peripheral therapy [5]. Over-the-needle catheters have a plastic cannula that fits snuggly over an introducing needle. Once the needle is positioned in the vein, it is removed, leaving only the flexible catheter.

Temperature of blood: Excessive heating of blood components alters the red cell membrane, which in turn affects the elasticity, deformability, and osmotic fragility of the RBCs [2]. The damage caused by excessive heating will depend on the exposure time, as well as the temperature. It is recommended that tubing is covered with aluminum foil or routed in such a way to avoid exposure to potential heat sources. The blood can also cool substantially before reaching the patient, especially when the blood is flowing at slow flow rates. Cold blood infusion risks hypothermia in neonates; therefore, the blood must be regulated at an optimal temperature.

Existing neonatal blood transfusion systems: There is no exact method for performing a neonatal blood transfusion; however, there are various existing mechanisms that assist in the blood transfusion method.

Electromechanical pump systems: Electromechanical pump systems are used to deliver cellular blood components, intravenous fluids, and medications [2]. Electromechanical pump systems provide accuracy in the volume to be delivered and the flow rate required. Neonatal electromechanical pump systems are more accurate in the short term (up to 60 minutes) when infusing at rates less than 5 mL/hours and the occlusion pressure is set lower than 300 mmHg).

Pumps: There are various types of pumps available that utilize different mechanisms of delivery in order to distribute blood to a patient. A peristaltic mechanism squeezes the administration set tubing to create flow [2]. The peristaltic-type pump produces a great deal of red blood cell damage; therefore, it is not recommended for use. A piston-actuated syringe-type mechanism uses a constant high speed refill cycle of the syringe followed by a slow ejection phase controlled by a check valve. A piston-actuated diaphragm cassette mechanism works by having blood fed into the cassette by gravity, where the flow is then controlled by two pistons acting against two check valves. The diaphragm cassette mechanism also produces a great deal of hemolysis; therefore, it is not recommended for use.

Blood Transfusions in Developing Countries

Blood transfusion practices in developing countries vary greatly from the blood transfusion practices of developed countries. The greatest differences in practices are due to the amount of material resources available to operate blood transfusion systems, the distribution of diseases and the treatment. In developing countries, pediatric blood transfusions make up a large percentage of transfused patients, specifically in subSaharan Africa. As a result of these challenges, blood transfusion services among resource-restricted countries vary considerably in their levels of development, safety, and quality.

Conditions requiring transfusions: The majority of neonatal transfusions are given for basic, usually urgent or life-threatening conditions, rather than for support of tertiary care needs [2]. Major causes of pediatric anemia in developing countries consist of the following: malaria infection, nutritional anemia due to inadequate dietary iron, thalassemia, hemoglobinopathy, HIV infection, chronic or recurrent infections, and accidents.

Malarial anemia: Malaria causes anemia primarily through repeated episodes of hemolysis [2]. Anemia can become quite severe in young children who have not yet achieved a degree of immunity against malaria. In developing countries of subSaharan Africa and parts of Asia, malaria is a primary cause for pediatric hospitalization due to severe anemia.

Nutritional anemia: Iron deficiency anemia is highly prevalent in many developing countries due to inadequate dietary intake of vitamin B_{12} or folate [2]. The WHO estimates that 50 % of children in developing countries are suffering from iron-deficiency anemia.

Thalassemia: Thalassemia is a genetic autosomal recessive blood disease, which results in reduced rates of synthesis of one of the globin chains that make up hemoglobin, and this in turn causes anemia [2]. The treatment of thalassemia places major demands on the blood supply in developing countries, especially in Southeast Asia, Africa, and the Middle East.

Hemoglobinopathy: Sickle-cell disease is one of the most common hemoglobinopathies [2]. Over 80% of the world's population with sickle cell disease is born in Africa; however, transfusions in pediatric sickle-cell disease patients are reserved for those with severe anemia (hemoglobin level less than 5 g/dL). Unfortunately, the regions of Africa where the hemoglobin gene is most common coincide with those where the HIV pandemic is greatest; therefore, patients with sickle-cell disease are recognized as a group at risk of acquiring HIV infection.

Material resources: Many developing countries have a short supply of the material resources needed to perform blood transfusions [2]. Throughout developing countries, there is a chronic shortage of blood products. This shortage has intensified by the problem of inappropriate transfusion practice and the failure to use sound medical guidelines for transfusion. It has been estimated that 13% to 47% of all pediatric transfusions in Africa are given unnecessarily. The blood inventory in many developing countries consists of only 1 or 2 days' supply; therefore, more often than not, the patient's family must locate a suitable donor to supply blood.

Whole blood, usually collected in CPDA-1 anticoagulant, is the most commonly available blood product [2]. PRBCs are frequently unavailable in developing countries, while FFP and platelets are similarly scarce in many resource-restricted countries. Private hospitals and major teaching hospitals may have PRBCs, FFP, and platelets for their use but typically only in short supply.

Small volume transfusions are sometimes performed by removing aliquots of blood into small volume bags, sterile syringe sets, or buret sets when available [2]. Infusions pumps are not widely available; therefore, infusions rates are typically determined by the drip-rate method. Various developing countries lack reliable sources of electricity and refrigeration, which has hindered the establishment of good blood transfusion methods.

Risks of infectious disease transmission by transfusion: In the year 2000, 31% of all donated blood units were not screened for one or more of the three most serious transfusion-transmitted viruses: HIV, HBV, and HCV [2]. Almost all of these screening lapses occurred in the world's developing world. Transfusion-transmitted HIV infection is one of the most serious adverse consequences of blood transfusion in children in developing countries. Approximately 10% to 15% of HIV transmissions in subSaharan African are attributed to transfusions; however, the proportion of infections attributable to transfusions in pediatric populations is actually much higher, as over half of transfusions are administered to children.

Guidelines for neonatal blood transfusions: The guidelines for neonatal blood transfusions are similar to those in developed countries; however, the guidelines tend to be more conservative due to the increased risk of adverse affects. Due to the high risk of transfusion-transmitted infectious disease and acute blood-product shortage in many developing countries, the guidelines listed in Table 4 are typically followed to determine if a neonate needs a blood transfusion (p. 10). Blood transfusions for neonates with a hemoglobin level less than 5 g/dL are only considered necessary if the neonate is experiencing respiratory distress or cardiac failure; however, if the neonate is clinically stable, it is advised that the patient is closely monitored and treated for the cause of anemia [2]. If a neonate acquires a hemoglobin level greater than or equal to 5 g/dL, then a transfusion is usually not necessary. Transfusion is only considered in cases of shock or severe burns when a neonate has hemoglobin level that exceeds 5 g/dL.

Table 4. Typical guidelines for pediatric7 transfusion in developing countries [2].

Hemoglobin Level	Patient's Clinical Condition
Less than 4 g/dL	----
Less than 5 g/dL	Respiratory distress or cardiac failure are present

Transfusions to neonates must be administered slowly, especially when whole blood is used; otherwise, the neonate may experience volume overload [2]. Whole blood transfusions are often administered at a dose of 20 mL/kg over 2 to 4 hours. When PCRBs are available, they are typically given at a dose of 15 mL/kg.

Benchmarking

The following section will describe various transfusion devices that are currently employed in the hospital setting, as well as various transfusion devices that have been patented by the United States.

Four-Way Stopcock

The Holden Neonatal Intensive Care Unit (NICU) at Mott Hospital in Ann Arbor, Michigan presently uses a manual ET process for neonates (Robin Jahnke, personal interview, February 11, 2009). The components needed for the manual ET process are provided in an ET Tray, which consists of two 20 mL syringes, an extension tube, a 5 Fr and a 8 Fr umbilical catheter, a blood administration set, and a fluid collection bag, which is illustrated in Figure 2 below. This device works much like the process presently used at KATH; however, the four-way stopcock excludes the necessity of having multiple valves. The device functions by rotating a single valve, which opens and closes the various pathways.

Figure 2. Manual four-way stopcock device presently used at various hospitals.

DRE SP1500 Plus Syringe Pump

In 2008, the U.S. Food and Drug Administration (FDA) approved the use of syringe pumps in medical facilities. A syringe pump provides accurate continuous infusion of medications at expanded flow rates. Syringe pumps can typically provide infusion via intravenous, intra-arterial, epidural and subcutaneous routes of administration.

DRE Inc. recently introduced the DRE SP1500 Plus syringe pump into the medical field, which is illustrated in Figure 3 below [11]. The DRE SP1500 Plus can infuse medications at rates ranging from 0.1 mL/hr to 1500 mL/hr. It also supports all major syringe brands with volumes ranging from 10 mL to 100

mL; however, the device allows multiple syringes to be stacked thus providing a no volume limit. The SP1500 Plus syringe pump allows physicians to quickly and simply enter parameters, such as the infusion rate limit and the total infused volume, which must be taken into account during the infusion procedure. The SP1500 Plus has the capabilities of operating both manually and automatically; therefore, allowing the physician to interrupt the process if needed and perform the process manually, or the physician may care for other patients while the infusion process is taking place. The SP1500 Plus runs off of a NiMH rechargeable battery. It is easily transportable for the device is 135 mm by 305 mm by 195 mm and only weighs 2.6 kg. Such syringe pumps cost between $495 and $1500.


Figure 3. DRE SP1500 Plus Syringe Pump [11].

Although such a device provides various advantages to infusing products, such as blood, the device only takes into account infusing blood and not ejecting blood from a patient. The device is also extremely expensive.

Exchange Transfusion System (Out-of-Phase)

Patent 4,457,747 introduces an automated ET device, particularly for newly born babies or children [12]. The device is depicted in Figure 4 (p. 12). This device uses two coupled, automatically driven syringes, one in a blood withdrawal system and one in a fresh blood injection system. The coupled system insures that the volume removed from a baby in the withdrawal system will be simultaneously replaced by an equal volume of fresh blood from the injection system. As depicted in Figure 4 (p. 12), the syringes are out-of-phase of each other, which entails that when the syringe on the left is injecting fresh blood into the neonate, the syringe on the right side is ejecting blood from the neonate. When the syringe on the left is filling up with fresh blood, the syringe on the right is emptying the waste blood into a waste container; therefore, the neonate is neither receiving nor disbursing blood during the refill phase. This device is run on an electric motor, which is connected to a box panel, depicted in Figure 5 (p. 12). The electric motor has appropriate gears to reduce the volume flow rate as desired. Although this device has various advantages, having two insertion points into the infant (the umbilical vein and a peripheral vein) is not plausible. The injection and ejection of blood from a neonate must be within a large blood vessel; otherwise, the injection and ejection of blood in a peripheral vein will cause the vein to collapse (R. Jahnke, personal interview, February 11, 2009).
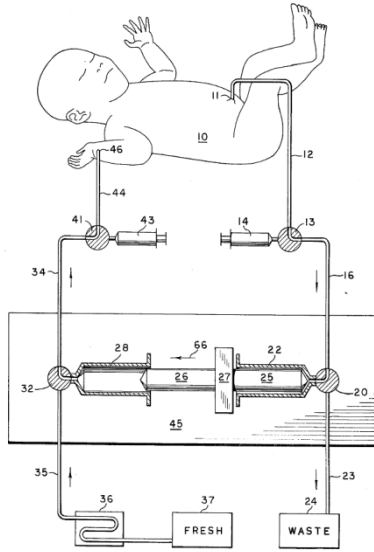
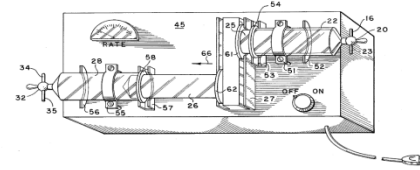**Figure 4. Automated exchange transfusion device [11].**

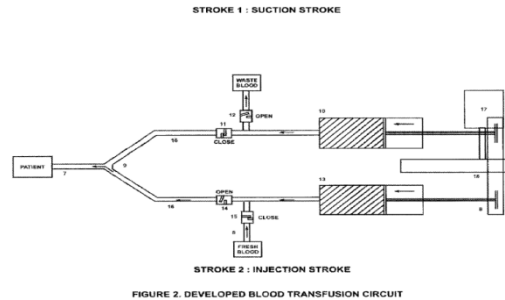**Figure 5. Box panel holding syringes, which are connected by a slider [11].**

Blood Transfusion System (In-Phase)

Patent 7,083,587 is very similar to that of Patent 4,457,747 described above; however, the automated ET device consists of an in-phase system, as well as a single injection point [13]. The device is depicted in Figure 6 below. A Y-shaped connector is used; therefore, the patient only needs one injection point. At one end of the Y-shaped junction is a waste blood syringe, while at the other end of the Y-shaped junction there is a fresh blood supplying syringe. The system functions by having motor push and pull a slider, which operates the syringes. First the syringe is pulled outwards; therefore, the top syringe ejects blood from the neonate, while the bottom syringe fills up with fresh blood. The suction process is depicted in Figure 6 below. When the syringe is then pushed forward, the top syringe empties the waste blood into a waste blood container and the bottom syringe injects the fresh blood into the neonate. The injection stroke is depicted in Figure 7 (p. 13).
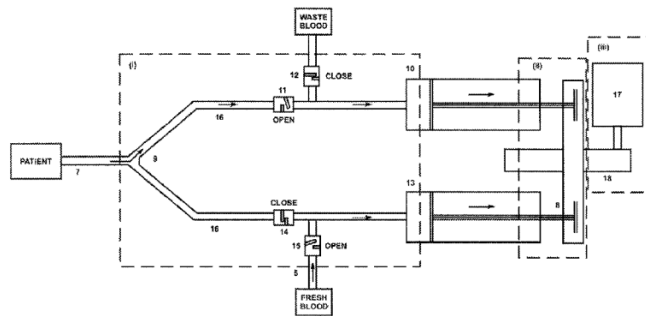
Figure 6. Suction stroke process of in-phase blood transfusion device. [13].

This patented design was created for the following reason: to replace a time-consuming, manual ET procedure with a system that can operate without the constant supervision of a nurse or doctor. This embodies the very objectives that KATH has bestowed upon us.

STROKE 1 : SUCTION STROKE

WASTE BLOOD

OPEN

CLOSE

PATIENT

OPEN

CLOSE

FRESH BLOOD

STROKE 2 : INJECTION STROKE

FIGURE 2. DEVELOPED BLOOD TRANSFUSION CIRCUIT

Figure 7. Injection stroke process of in-phase blood transfusion device. [13].

Information Sources

To familiarize ourselves with the ET process and equipment, the precautions and risks in caring for neonates, and pre-existing ET device designs, we consulted a variety of sources and mediums of information. As a starting point, we utilized internet search engines for general knowledge. As we researched more into our specifications, we resorted to more reliable sources of information; namely medical journals and books, United States patents, and interviews with pediatric professionals.

We found many great sources to obtain coarse information about the topics regarding our project. Medical journals and books were our first source of bulk information past the preliminary knowledge gathering stage. A combination of PubMed and Google led us to *Pediatrics: The Official Journal of the American Academy*. We obtained enough information from these journals to begin a list of specifications for our device. Newborn blood volumes, blood flow rates for different weight classes of neonates, and risks such as air embolism, were among the specifics we acquired from the journal. These are all engineering considerations we must take into account for our design. Another medical book which included the procedure for a manual ET was The Clinical Use of Blood by the World Health Organization.

With a solid understanding—gained from medical journals, books, and other supporting documentation—of ET and its related topics, we were prepared enough to interview local pediatric professionals. This phase of knowledge gathering was the most efficient medium and provided the most information regarding medical factors that we need to keep in mind, such as max blood flow rates and possible air embolisms.

Project Requirements and Engineering Specifications

After conducting numerous hours of background research to learn as much as possible about ETs, we were ready to develop a list of qualitative customer needs that would be essential to the success of our automated ET device. To advance our goals, each customer need was quantified in some way to create an engineering target or specification that we will be trying to reach during the design process. Medical journals, United States patents, medical personnel at Mott's Children Hospital, and Dr. John Adabie, our mentor from KATH, were the key resources we used to help quantify our customer needs. Once we derived our engineering specifications, we created a QFD diagram (Appendix A-2) to map the customer needs against the engineering specifications to quantifiably determine which specifications would be most important to address. We used a 1, 3, 9 scale to indicate weak to strong relationships respectively between the customer needs and the engineering specifications. Table 5 summarizes these relationships (p. 15). (Table 5 includes updated customer requirements and specifications in bold, underlined black font.) The specifications we deemed most important are explained individually below.

Necessity of Medical Personnel during ET Process

The overlying customer need and main motivation behind designing an automated ET device is to increase the number of available medical personnel; therefore, providing doctors and nurses with more time to care for other patients. The device is meant to replace the manual ET procedure that is currently used at KATH through means of being automatic. Being such a broad customer need, all that can be said about it so far is that the device will operate with synchronous intrusion/extrusion strokes, and it will ultimately rely on the success of meeting the other customer needs, such as the ease of use, ease of assembly, and ease of mobility.

Inexpensive

Funding is extremely limited in the medical field in Ghana; therefore, the ET device needs to be as inexpensive as possible. We are targeting a cost of US $200 or less per device (all parts included). Automated infusion pumps, an existing medical device used at various U.S. hospitals, costs as little as $495 and as high as $1500. Much of this expense is due to the high technology incorporated with the design. The use of disposable parts presently available at KATH will help drive the cost down of each device. The main concern for keeping the cost minimal is the cost of the driving mechanism used to run the device. The driving mechanism was chosen to be a linear actuator, and this will be the most expensive component of the automated ET design.

Ease of Use

The device must be easily employed by medical personnel. The device will be automated; however, that does not necessarily mean that it is easy to operate. The ET device must have a self explanatory user interface. Ideally, the start-up process should be the only time a doctor or nurse is actually using the device other than when it shuts off at the end of the transfusion. Although the automated ET device will function with limited interaction from the medical personnel, the neonates must be intermittently monitored by medical personnel to examine the neonate's vitals.

To begin the transfusion process, the volume flow rate will need to be set to between 0 and 10 mL/min according to the size of the neonate, and then the motor mechanism will need to be started. The device should not require any form of human interaction beyond that point until the end of the transfusion, outside of occasional check-ups to make sure there have been no malfunctions. An easy-to-use device, where the process does not require medical personnel beyond the start and end of the transfusion process, will offer medical personnel with more time to care for other patients in the MBU.

Automatic Shut off

An automated shutoff system is essential in the design of an automated ET device. If the system does not automatically turn off, the neonate will be at greater risk of experiencing volume overload. To insure the safety of the neonate, a feedback system will need to be implemented. The feedback system will detect the amount of blood administered into and out of the neonate. When the system has reached the indicated blood volume, the feedback system will ignite an automatic shut off. The microcontroller within the circuit board will be responsible for shutting down the process once the desired total volume of blood has been transfused.

Prevents Air Bubbles

One of the biggest risks when using a hypodermic on a patient is the transmission of air bubbles from the syringe into the circulatory stream. The topic of air embolism was discussed previously in the "Risks associated with neonatal blood transfusion" section above (p. 6). The automated ET device must insure the safety of the neonates by means of preventing air bubbles from forming in the syringe(s). A specific procedure needs to be conducted to eliminate all air bubbles before the automatic ET process begins. This

procedure will entail 'priming' the tubing, or filling the tubing with blood before turning the system on. This will eliminate the possibility of air bubbles forming in the tubing or the syringes. The procedure will be designed to reduce the volume of air in the syringe to less than 0.1 mL. A bubble as small as 10 mL in the heart may cause hypoxia, and a bubble only a fraction of a milliliter located in the brain may cause a stroke. Therefore, this design criterion is one of primary importance.

Ease of Assembly/Disassembly

The device will have many parts, but the assembly process should not be laborious. It should take no more than 10 minutes to assemble/disassemble the device. The device should come with a clear, one-page instruction manual that makes assembly as easy as possible. The device will be made with available parts already at KATH; therefore, the staff at KATH should be familiar with most of the components already, creating an easier assembly process. Also, the motor/linear actuator should not weigh more than 10 lbs (4.54 kg), so picking it up for assembly is not an issue for any doctor or nurse.

Functions with Battery Power

AC power is not the most reliable source at KATH given that the power goes out occasionally. A dependable power source is needed to operate the ET device, especially because there is a life on the other end of this device. If the device fails, the infant is at a greater risk of dying. In order to take advantage of the AC power, but also insure a reliable electrical source, a battery operated system should be implemented. This system may also employ an AC power system, while using the battery operated system as a backup. In order to implement a battery into the ET design, the battery must be readily available in Ghana. The battery must also supply enough power to operate the entire system. Due to such criteria, the supplied battery power should be no more that the power emitted by 4 standard D sized batteries (1.5 Volt, 4500mAh) in series, which will need to cost no more than US $0.50/battery.

Applicable with Various Syringe Sizes and Brands

Syringes come in various geometries. Since a syringe is being used as the pumping mechanism in the ET device, it is crucial to design a system that is compatible with various types of syringes. Syringes do not only have various geometries, but they also have various properties, such as the frictional force between the plunger and the syringe. Ideally, the ET device should accommodate syringes that range from 20-25 mL because a neonate should not have more than 25 mL injected or ejected in a single stroke. To create a system that accommodates such a wide range of syringe volumes and types, there will ideally need to be a feedback system to control the pumping of the syringes.

Minimizes Disease Transmission
The automated ET device will be dealing with blood transfers; therefore, the need for this device to prevent disease transmission is vital to insure the safety of the neonates. In order to meet this requirement, our device will have to meet the Ghanaian health regulations. Also, to enhance the probability of preventing disease, our device will consist of approximately 50% disposable materials.

Instruction Manual

The device should come with a short, less than five-page instruction manual that clearly explains how to use it properly. We are trying to develop a simple design that will have a high success rate; therefore, a complex

Table 5. Engineering specifications and target values to meet customer requirements.

Customer Needs	Customer Weight (9=High Importance 3=Medium 1=Low)	Technical Requirement	Target Value

15

Constant monitoring from doctor is not required	9	Synchronous intrusion/extrusion strokes (mL/stroke)	25
Inexpensive	**3**	Total cost (USD)	≤ 200
Ease of use	9	Operating/assembly instructions (pages)	≤ 1
Automatic shutoff	9	Feedback system	Yes
Prevents air bubbles	9	Volume of air bubbles pre-process (mL)	≤ 0.1
Ease of assembly/disassembly	9	Setup time (minutes)	≤ 5
Functions with battery power	**3**	Battery replacement time (days)	≤ 7
Applicable with various syringe sizes and brands	9	Compatible syringe sizes (mL) Feedback system	20-25 Yes
Prevents disease transmission	9	Disposable Components (% of total)	50
Instruction manual	3	Operating/Assembly Instructions (pages)	≤ 5
Adjustable volume flow rate	**9**	Volume Flow Rate (mL/min)	0-10
Transfuse various types of blood (PRBC, FFP, platelets, etc.)	3	Volume Flow Rate (mL/min)	0-10
Easily maintained and repaired	3	Disposable Components (% of total)	80
Utilizes available spare parts	3	Spare Part Replacement Time (days)	≤ 7
Operable for both blood transfusion and exchange transfusion	3	Number of insertion points	1
Mobile	3	Total Weight (N)	**≤ 65**
Blood volume control	3	Total blood transfused (mL/kg) Feedback system	10 Yes
Safety of neonate	3	Motor Noise Level (dB) IV insertion point	≤ 45 1
Functions with AC power	**9**	Operating Voltage/Frequency (V, Hz)	220, 50
Easy transition to manual process	1	Time to transition from automatic process to manual process (seconds)	< 60
Minimize the size	1	Volume Dimensions (m³)	6688
Ease of sanitizing/cleaning	1	Disposable Components (% of total)	50
Minimize sound	1	Motor Noise Level (dB)	≤ 45
Alarm device to detect malfunction	1	Alarm Noise Level (dB)	70
Alarm device to detect the end of the process	1	Alarm Noise Level (dB)	60

instruction manual should not be necessary. Along with written instructions, the instructions manual should have a schematic diagram of the device so the procedure is easy to follow visually. Any instructions manual longer than five pages in length will require unnecessary time spent reading, and that is unacceptable as our main goal of this project is to open up more time for the medical staff so more babies can be cared for. A repair and maintenance instruction manual will be included as well to assist the staff at KATH make any repairs that may be needed throughout the lifetime of the device.

Adjustable Volume Flow Rate

The volume flow rate in which blood can be safely infused and ejected from a neonate is a critical specification. A suitable volume flow rate of blood flowing into and out of a neonate is 3 mL/min [3]; however, the volume flow rate may need to be increased or decreased depending on the type of blood being infused into the neonate due to the varying fluid viscosities of different blood components. In order to prevent hemolysis of blood components due to mechanical damage from rapid infusions through the small-gauge needles that are needed for neonates (<24 gauge), a low volume flow rate (3 mL/min) must be utilized [3]. The neonate is also at risk of having a volume overflow, which is often caused by a rapid volume flow rate. The automated ET device must have an adjustable flow rate. Although neonates typically receive blood at 3 mL/min, we would like to create a device that operates at 0-10 mL/min to incorporate any need for a larger volume flow rate.

Transfuse Various Blood Components

KATH transfuses various types of blood components into neonates, such as whole blood, PRBCs, FFP, and platelets. Whole blood and PRBCs are the primary blood component that is administered into neonates; however, these two blood components have varying fluid properties. Due to the varying fluid properties of the blood, the volume flow rate required will also vary.

Easily Maintained and Repaired

The ET device needs to be easily maintained and repaired. 90% of medical supplies shipped to foreign countries like Ghana do not arrive in working condition (B. Teninty, personal interview, January 20, 2009). The automated ET device should be easy to repair; therefore, the device does not merely prop a door open, but rather maintains its original function. A majority of the parts will be disposable, such as the plastic tubing, syringes, and valves. Such parts should be replenished within a week's notice if they ever run out. If other parts such as the linear actuator or power supply are damaged in the shipping process, the one-page instructions manual should signify how to fix the issue. An automated ET device will be extremely beneficial to KATH; therefore, we must insure that the device is easily repaired.

Utilizes Available Spare Parts

A device that utilizes spare parts currently available at KATH will play a major role in the design process. If many of the components presently at KATH are utilized then the less the hospital will have to spend on importing equipment. When spare parts run out, KATH should be able to receive new parts within one week's time.

Operable for Both Blood Transfusions and Exchange Transfusions

The automated ET device must also be able to function as a blood transfusion device; therefore, blood will no longer be ejected from the neonate, but only administered into the neonate. A blood transfusion is often required is a neonate is hemorrhaging, in which the blood must be administered as quickly as possible in order to replenish the neonate's blood supply. By only having one insertion point, it will create an easier transition between the ET process and the blood transfusion process. A tube or valve may need to be removed from the system; however, this is a design based specification.

Ease of Mobility

The device must be easily transported from ward to ward, and incubator to incubator. Even though the neonatal ETs at KATH take place on open tables, we intend to make our device capable of fitting easily into an incubator occupying an infant. A normal incubator box is approximately 19×40×22 inches (0.48×1.02×0.56 m), however this must contain both the baby and the device comfortably. To insure plenty of room for the neonate, the ET device should not consume more than one third of the incubator space (5573 in^3 or 0.091 m^3). The relatively small device volume will assist in making the device more mobile. The device shall also not weigh more 15 lbs (4.54 kg). In order to aid the medical personnel in transportation of the ET device, the device needs to be picked up with a minimum amount of work. The minimized weight and size values will make the device easier to move.

Blood Volume Control

An automated process does not require medical personnel to constantly monitor the neonate's progress; therefore, the doctors and nurses will not be aware of the amount of blood that has been transfused into and out of the neonate. The device must be able to detect the amount of blood that has been injected into and ejected from the neonate. The total blood volume inputted into the neonate is typically 10 mL/kg of the neonate. The volume of blood entering and exiting the neonate will be calculated by the number of cycles that occur within time of operation. This number will then be an input into a feedback system, which will cause the system to stop after the correct amount of blood has been transfused.

Safety of the Neonate

There are many safety considerations to take into account when dealing with ETs in neonates. ETs must take place in fairly large blood vessels, which becomes a fairly large concern considering that neonates have very small blood vessels. Due to the small blood vessels of neonates, ETs are limited primarily to the larger blood vessels, such as the umbilical vein. The umbilical vein is the best blood vessel to use for an ET, because it is larger in size and does not require surgery to access. Due to the limited number of large blood vessels in neonates, the ET process should only require one IV insertion point.

The motorized device will be next to the neonate during operation, because tube length between the insertion point and the syringes must be kept at a minimum to reduce flow energy losses. Due to the close proximity between the device and the baby, the device should not operate at a noise level greater than 45 dB, which is roughly equivalent to a refrigerator hum.

The automated ET device should also be constrained to the volume dimensions stated in the ease of mobility section above. As stated, the neonate and the device will share the same tabletop or incubator. We want the neonate to have sufficient space to move in without coming into contact with the device.

Functions with AC Power

The most convenient power source available would be an AC power source (with the exception of the power occasionally going out). This would supply the device with power that does not waste one's energy, especially through means of a mechanical powering system. For AC power to be a viable means of powering the ET device, the device will need to be compatible with a 220 V, 50 Hz electrical supply.

Easy Transition to Manual Process

The automated ET device will consist of many of the same components currently used at KATH to insure that the ET device can easily transition from an automated process to a manual process. The device must allow the medical personnel to easily stop the process and insert a syringe into the system in which they can eject and inject blood manually. It should take less than 60 seconds to transition from the automated

process to the manual process. This is an essential feature, especially if the device is not working as desired.

Minimize the Size

The size of the device needs to be small enough to be moveable, as well as to be operable in the environment needed. The environment will be the MBU, so a preferred specification would be to have the device operable inside an incubator. This would only be feasible if the neonate was also able to fit comfortably inside the incubator. A standard incubator is 19x40x22 inches, while an infant comfortably consumes approximately 19x24x22 inches; therefore, the automated ET system needs to fit within a volume of 19x16x22 inches.

Ease of sanitizing/cleaning

The automated ET device will primarily consist of disposable parts presently available at KATH; however, approximately 50% of the system components will not be disposable. Due to use of blood components within the ET system, sterilization is necessary to eliminate the risk of transmitting infections. To insure a quick and easy sanitizing process, there will be a minimal number of non-disposal products. The components that are not disposable but come in contact with blood will be sanitized using a bleach-water solution.

Minimize Sound

In a hospital, extraneous noise can be very distracting if not dangerous. Due to this, the regular running of our device needs to be quiet. A maximum dB level of the device during regular operating system can be no more than 30dB.

Alarm Device to Detect Malfunction

The device is intended to be automatic and functional without the presence of medical personnel. The device should be equipped with an alarm system that will activate upon malfunction of the device or any sort of deviation from the ET process. This is a safeguard against any sort of emergency related to the malfunction of the device. If the device locks up before the end of process, if the blood flow rate changes from the initial setting, or any other numerous malfunctions, an audio alarm of 70 ± 5 dB will activate notifying personnel immediately. 70 ± 5 dB is roughly the same magnitude of an alarm clock. It would be audible from anywhere in the ward—even if music is being played on the radio, which according to first-hand accounts is sometimes the case. Hearing loss for a sustained exposure does not occur until approximately 90 ± 5 dB. The alarm would potentially be the same sound for any type of malfunction, but the requirement of having an alarm device to detect malfunction is no longer of high priority for a feedback system will be implemented to insure the proper volume flow rate.

Alarm Device to Detect the End of Process

Once again, since the device is intended to be automatic and functional with personnel being present, the device will be equipped with an alarm system that will activate upon the end of the ET process. This alarm will be quieter and less striking than the alarm for a malfunction. It will be an audio alarm of 60 ± 5 dB—the equivalent of an electric shaver or the strike of a piano key. The alarm is intended to notify personnel that the process has completed; however, this alarm has less urgency than the malfunction alarm, because our device will be designed to cease operation at the end of a pre-determined process anyway. The 60 ± 5 dB sounds does not come within the sustained 90 ± 5 dB range needed for hearing loss. The requirement of having an alarm device to detect the end of process is no longer of high priority for a feedback system will be implemented to insure the process ends after the neonate has been infused with the indicated amount of blood.

Concept Generation of Full Systems

The brainstorming process is a key factor to developing a successful prototype; therefore, various brainstorming techniques were employed during the concept generation phase. From the brainstorming process, five main concept designs were formulated. The five best concepts are discussed in detail in the following sections.

Brainstorming

Before the brainstorming process could take place, it was essential to develop a functional decomposition diagram (provided in Appendix A-3) outlining the inputs and outputs of our device. This assisted us in determining the key components and subsystems to highlight in our brainstorming process.

Before coming together to brainstorm ideas for design concepts as a group, we utilized an individual concept generation process to come up with as many ideas as we could on our own. For this part of the process, we focused on developing whole systems. The focus was to not spend too much time worrying about what may or may not work with the systems, but rather just to get as many ideas as possible written down, regardless of how impractical or improbable they would be to implement. Each team member was expected to develop between 3 and 4 whole systems prior to coming together for group brainstorming.

Once our group had developed concept ideas individually, we came together to discuss our ideas and to brainstorm new ones based on what we had developed already. Also for this meeting, we broke down our developed whole systems into subsystems to brainstorm different ideas for some key components of our system (i.e. power source, valve control, and insertion possibilities). For this brainstorming session, similar to the individual process, we did not worry about developing practical concepts. We merely wanted to develop as many systems as we could to have as many ideas as we could.

We began the group session discussing the ideas we had come up with individually. We then moved onto developing new systems that had sprouted from our individual ideas, as well as developing new ideas for our subsystem components. We chose to separate our whole system concepts into categories of how they would be powered (i.e. electronically, mechanically, or by gravity). Once we had derived as many ideas as we could on a large chalk board, we copied all of our concept ideas onto small Post-Its for recording purposes. (The various design concepts that were formed during the brainstorming process are provided in Appendix A-4.)

After our group concept generation session, we displayed our ideas to our discussion section in the form of an informal presentation to get some peer feedback for further brainstorming. From this session, we got some valuable feedback that helped solidify some ideas we already had, and some new ideas that would help advance some of our concept generations.

The next step was to put together all of the brainstorming we had done so far, combine ideas and components, and come up with the best 5 concepts to consider for a final concept selection. Once we chose our top 5 whole systems, we developed them further on dry erase boards to have a visual idea of how the entire system might look if created. The five concepts we chose to decide between for final concept selection are explained in the following sections.

Out-of-Phase Slider

One concept that was developed during the brainstorming process is referred to as the *out-of-phase slider*, which is depicted in Figure 8 (p. 21). The out-of-phase slider system consists of a driving mechanism and a pumping mechanism. The driving mechanism is revealed as a motor-crank system in Figure 8; however, there is much flexibility in regards to the design of the driving mechanism, especially since the primary

focus of the driving mechanism is to power the pumping mechanism. The driving mechanism, in this case a motor, crank-connecting rod system, applies a force onto a slider. The slider is connected to two syringes, in which one syringe is located to the left of the slider and the other syringe is located to the right of the slider. This slider-syringe component acts as the pumping mechanism.

The pumping mechanism allows the syringe to act as a pump, in which the slider will push the plunger of the syringe back and forth; therefore, allowing the syringe to fill up with blood, as well as empty blood. The syringes are then connected to the neonate via a catheter and some tubing. Figure 8 below illustrates that there is also a junction intersecting the connection between the syringe the neonate. Each junction is connected to a reservoir, which either supplies fresh blood to the system or serves as a waste blood container. Each junction consists of a check (one-way) valves. A check valve insures that the blood can only flood in one direction. The check valve located at the junction near the fresh blood will insure that blood can only flow out of the reservoir. The check valve will also serve the purpose of preventing blood flowing from the neonate towards the syringe. The check valve located at the junction near the waste blood container insures that waste blood can only flow into the waste blood container; therefore, waste blood cannot exit the waste blood container and reenter the neonate.
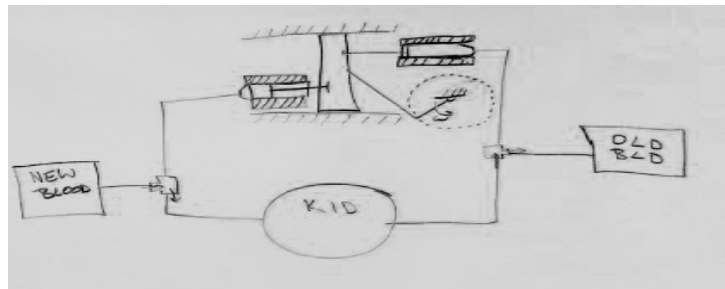
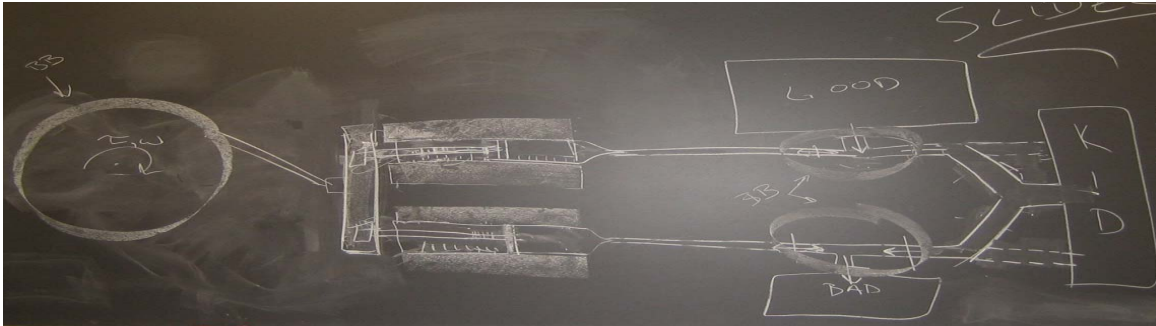**Figure 8. Out-of-phase slider design concept schematic.**

The pumping mechanism process has the two syringes pumping out of phase; therefore, when the syringe on the right is ejecting blood from patient, the syringe on the left hand side is simultaneously injecting blood into the patient. Similarly, as the syringe on the right is pumping waste blood into the waste blood container, the syringe on the left is being filled with fresh blood. This system provides the neonate with a stable blood pressure for the loss of blood is immediately replenished, which is due to the two insertion points. This design also provides a break in which blood is neither being injected into nor ejected from the neonate.

In-Phase Slider

One concept that was developed during the brainstorming process was the *in-phase slider*. The in-phase slider is very similar to the out-of-phase slider; however, the two syringes pump in-phase with each other. The in-phase process indicates that each syringe simultaneously fills with blood and simultaneously empties of blood. In order to create an in-phase system rather that an out-of-phase system, the two syringes must remain on the same side of the slider, as depicted in Figure 9 (p. 22).

The syringes are connected to the neonate similarly to the out-of-phase slider system; however, a Y-shaped junction connects the two syringes to a single injection point. It is plausible to connect the two syringes together via a Y-shaped junction because blood is not being injected into and ejected from the neonate simultaneously. Rather, the syringe located on the bottom of Figure 9 (p. 22) will eject blood from the neonate when the slider is forced away from the syringes, while the syringe on top will be ejecting blood from the fresh blood bag. The slider will then push forward towards the syringes, which will cause fresh blood to be injected into the neonate via the top syringe. This will also cause waste blood

to be injected into the waste blood container. The in-phase slider system also consists of check valves similar to that of the out-of-phase slider, which prevents blood from flowing in the incorrect direction. This system insures only one insertion point, which creates a much easier process for the medical personnel, especially since neonates have a minimal number of relatively large blood vessels. The process is constantly injecting blood into or ejecting blood out of the neonate.

Figure 9. In-phase slider design concept schematic.

Gravity-Piston

One concept derived during the brainstorming session is referred to as the *gravity-piston mechanism*. It is powered solely by the force of gravity and obviates the use of a motor. The system consists of a cylindrical tank of fresh-blood positioned above the neonate and an empty cylindrical tank with null pressure, positioned lower than the neonate and concentric with the fresh-blood tank as shown in Figure 10 below. Each cylinder is equipped with a piston initially at the top of each cylinder and the pistons are linked by a shaft. The neonate has two IV insertion points, one for insertion of blood and the other for extraction of blood. The insertion IV is connected to the higher tank of fresh-blood and the extraction IV is connected to the blood disposal tank below. The flow from the tank is blocked by a two-way valve until the process is ready to begin. A closed two-way valve is outside of the extraction IV until the process is ready to begin as well. The fresh-blood tank is filled with the amount of blood to be transferred into the neonate.
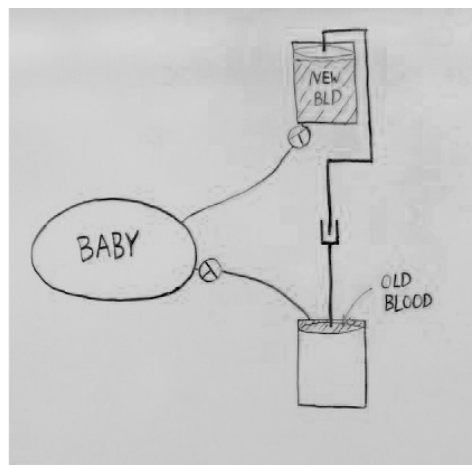

Figure 10. Gravity-piston device schematic.

The process begins when both valves are manually opened and blood naturally flows from the neonate into the disposal tank below which contains a pressure less than that of the neonate's circulatory pressure. The blood flows into the tank and onto the piston initially positioned at the top of the tank. The weight of the blood forces the piston down and forces the piston in the high tank down proportionally by means of the connecting shaft. This downward force by the piston through the force of gravity pushes the blood from the fresh-blood tank through the IV needle and into the neonate's circulatory system. To prevent the volume flow rate from exponentially increasing because of the increasing weight on the bottom piston a spring-damper system on the shaft was proposed to be implemented.

Compressor System

Another concept derived during the brainstorming session is the *compressor system*. It is powered by the force of gravity and the pressure differences between the neonate's circulatory system and blood reservoirs. The system consists of a bag of fresh-blood positioned above the neonate and a bag for waste-blood positioned below the neonate. The waste-blood bag is initially a vacuum which would induce a natural flow of blood out of the neonate and into the disposal bag. The pressure in the fresh-blood bag is initially higher than that of the neonate's circulatory system to induce a natural flow of blood out of the fresh-blood bag and into the neonate. In order for the child not to suffer from a net loss of blood, or from a vein rupture due to an excess of blood, the flow rates between the reservoirs of blood and the neonate will have to be equivalent. This implies that the pressure difference between the fresh blood reservoir and the neonate would have to equal the pressure difference between the neonate and the waste blood reservoir. To account for this, the waste-blood bag is equipped with a pressure sensor to detect the pressure inside the bag and to sends this information back to a microcontroller. The fresh-blood bag is placed in a sealed, rigid container that is equipped with a compressor that is linked to the microcontroller. There are also two-way valves outside of both the neonate and the fresh-blood bag to hold back flow until the process is ready to begin. Figure 11 below shows a schematic of the system.
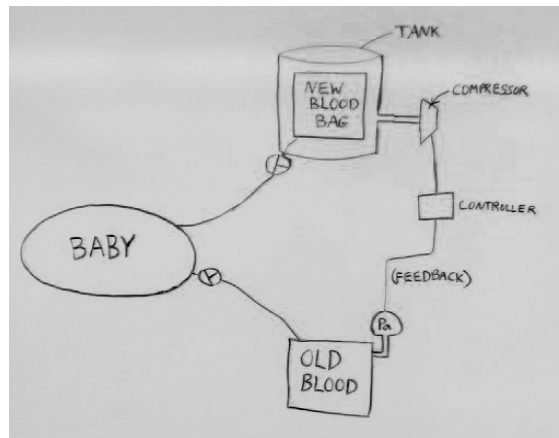
Figure 11. Compressor system schematic.

Automated Four-Way Stopcock

This system would incorporate the method currently used at the University of Michigan Hospital in the neonatal intensive care unit, with a few added features to make the system automated. With this four-way stopcock system, there are 3 tubes sprouting off the 3 open holes on the sides of the stop cock. One of the tubes is for removing blood from and injecting blood into the neonate, another is for removing blood from a fresh blood bag, and the last is for ejecting waste blood into the waste blood reservoir. The fourth part to the four-way system is the syringe that is on top of the stop cock that removes and injects the blood throughout the procedure. For the manual system, there is a valve that is on the stop cock that gets turned

by the doctor/nurse performing the ET when it is time for a new part of the process to take place. The concept we designed for this system would incorporate two motors to eliminate the human interaction. One motor would be a DC motor to power the pumping system, and the other would be a step motor to automatically turn the valve to different positions at the appropriate times. A visual of the manual process this system is based off of is provided in Figure 2 (p.10).

Concept Selection of Full System

After we developed our top five concepts into more complete designs, we needed to conclude which of the five systems satisfied the most customer requirements and engineering specifications. We had preliminary intuition that the in-phase slider mechanism would be the best choice of the five concepts, but we developed a Pugh chart to back up our intuition with quantified results, depicted in Table 6. Results from Pugh chart indicating that the in-phase slider is the best design concept.. We made a list of our most important customer requirements, and then compared how each system related to the current method being used at KATH for each requirement. If the system improved the customer requirement, a score of 1 was given for that category. If the system had a negative impact on the customer requirement, a score of -1 was given. If the difference was negligible, a score of 0 was given. The scores were added up for each system, and the Pugh chart results verified our intuition that the in-phase slider mechanism was the best design. (The entire Pugh chart, which includes the criteria in which we evaluated the five designs, is provided in Appendix A-5.)

Table 6. Results from Pugh chart indicating that the in-phase slider is the best design concept.

	Concept 1 *Out-of-Phase Slider*	Concept 2 *In Phase Slider*	Concept 3 *Gravity Piston*	Concept 4 *Compressor System*	Concept 5 *Automated Four-Way Stop Cock*
Σ+	+7	+7	+6	+6	+6
Σ-	-5	-4	-9	-10	-7
Σ_{TOT}	**2**	**3**	**-3**	**-4**	**-1**

The out-of-phase slider was a very similar design compared to the in-phase slider. The major negative impact of the out-of-phase slider, however, was the requirement of two insertion points. Based on our conversation with Robin Jahnke, a nurse from Mott hospital, it is essential that the transfusion system requires only one insertion point due to the difficulty of using any insertion point other than the umbilical vein in a neonate. Because of this issue, we decided against the out-of-phase slider.

The controller gravity system and the gravity piston system ranked as the worst designs quantitatively. They both were designed without a motor, so they fulfilled the inexpensive customer requirement compared to some of the other designs. However, both of these systems would have trouble keeping a consistent volume flow rate, and both would require medical personnel at KATH to transfer blood from the blood bags they are in currently to a special container, which would greatly inconvenience the KATH staff. This would also increase the amount of sterilizing that needs to be completed, which could increase the potential of transmitting infections.

The automatic four-way stop cock system would be a very easy-to-use system, but two motors would be necessary to operate the system. A stepper motor and some sort of servo motor would both be required to automate this system. With two motors, cost was a major issue for this system and that ultimately resulted in it not being chosen as our alpha design.

The in-phase slider improved on the most customer requirements compared to the negative impacts it would have. Some of the key customer requirements the in-phase slider satisfies are: it will allow for a controlled volume flow rate with the help of the feedback system, it will be applicable with various syringes, it only requires one insertion point, and it could be operable for both blood transfusions and exchange transfusions. The main downside of this system is the expense and possible maintenance of the high-tech electrical parts, but we feel the advantages significantly outweigh the disadvantages. Because of these reasons, we selected the in-phase slider to be our alpha design.

Sub-Function Concept Generation and Selection

After performing our first concept generation and selection processes to narrow down which full system would be selected as our prototype, we had to conduct a second concept generation process to decide which sub-functions would be used for the in-phase slider. A description of the sub-functions considered for each problem and the chosen solutions are as follows.

Driving Mechanism

When we first designed the in-phase slider, we envisioned a CAM system with a gear/crank and connecting rod to move the slider back and forth. However, there was a need for developing other options since we had not really explored alternative driving mechanisms. Also, we decided a CAM would be extremely impractical for a medical device such as the in-phase slider. After our brainstorming session with the discussion section, a method of using a belt or chain instead of the CAM mechanism was brought to our attention. A belt or chain would produce a more constant actuation than the CAM would, because it would be moving in a linear motion for the majority of its revolution as opposed to a circular motion. However, after further research into a possible belt or chain mechanism, we learned that belts are prone to a high failure rate and are difficult to maintain and repair. Because of these issues, we decided to again look for alternative methods. As we researched further, we found linear actuators online that perfectly performed the driving mechanism task that our device would require. A linear actuator moves in a back and forth, linear motion and would be able to drive our slider and the syringe plungers in and out, removing and inserting blood accordingly. Because we found linear actuators that would be more than capable of driving our slider for an affordable price, we selected a linear actuator to be our final driving mechanism.

Volume Flow Rate Detection

To detect volume flow rate in our system, we originally anticipated using a pressure transducer to detect the flow rate outside of the y-junction of our system, just before the blood would enter or just after the blood would leave the neonate. We chose this point because it would give the best feedback in terms of the actual flow rate entering the child. However, after questioning the feasibility of detecting flow rate inside the tubing using a pressure transducer, we decided to explore new options.

Choosing the linear actuator as our driving mechanism actually took care of this volume flow rate detection problem, which was another reason we chose it. The linear actuator contains a potentiometer, which is used as a position transducer to detect position. Instead of directly detecting volume flow rate, the potentiometer will constantly detect the position of the actuator. The microprocessor will then calculate the volume flow rate using the position of the actuator, the amount of blood that has been inserted or removed, and the cross-sectional area of the syringe being used. If the actuator ever reaches a position faster than it should, the actuation will either stop or slow down to allow the device to get back to its intended volume flow rate.

For this application, a pint of fresh blood should be sufficient enough for a neonatal ET; however, an ET requiring more than a pint of blood will require the ET device to be set to transfuse one pint of blood. After the one pint ET is complete, the medical personnel will then need to reset the ET device to transfuse

the remaining amount of blood needed. A current sensor would be ideal for this scenario for it would detect leaks, as well as when the fresh blood bag is nearing empty; however, due to the short timeline of this project, as well as the lack of funding, it is not incorporated into the design.

Slider

We generated numerous ideas for our slider before we reached a final decision. For the slider itself, we first created a CAD drawing of a block-shaped slider using CATIA. We then developed an appropriate holder that would allow the slider its necessary movement, and also hold the syringes in place so that only the syringe plungers would be able to move along with the slider. We first designed the holder to have circular grooves that the syringes would fit into, but upon discussing manufacturing and stability with Bob Coury, we changed the grooves to triangular slits.

A second generation process of the slider and holder brought us to the idea of purchasing a track used for opening and closing a desk drawer. Upon finding an inexpensive ball-bearing track made of stainless steel online, we decided to purchase that to allow the movement of the slider. The cost of purchasing the stainless steel track was a small amount greater than what manufacturing the block slider would have been. However, we concluded that the manufacturing time saved would be more beneficial to our team. Once we purchased the track, we made a new CAD of a simpler slider that will allow for the attachment of the syringes. We plan on using rapid prototyping to manufacture this slider.

The ball-bearing track will need to be protected from debris and fluids to insure that the track does not become corroded and rusted. To insure that the ball-bearing track is not impacted by debris and fluids, the track will need to be encased. The ET device including the linear actuator, syringe holder, and slider, will be encased by walls, which are described in detail in the "Fabrication" section below; however, the track may need its own encasing to prevent corrosion.

To hold the syringes in place in the grooves of the holder, we first developed the idea of using some kind of vice-grip mechanism. This would prevent unacceptable movement and rotation of the syringes, but the possibility of compressing and damaging the syringes if the vice-grip was tightened too much resulted in this method not being chosen. Using straps such as nylon backpack straps to hold the syringes in place was also an idea we generated, but the risk of slipping would be too great for a device such as ours. ¼" plate of Plexiglas will be used to hold down the syringes and prevent them from rotating and translating. Since various sized syringes will be applicable for this device, a porous material will be fixated to the plate of Plexiglas and to a piece of PVC. The porous material will allow syringes of various sizes to fit for the porous material will allow the top plate holding the syringes to adjust its height as needed. A butterfly nut will then be used to tighten the syringes in place as desired. The instructions for the device will direct the medical personnel to tighten the butterfly nuts so that a snug fit exists between the Plexiglas top place and the syringes. The butterfly nuts should not be overly tightened to prevent the PVC and Plexiglas from wear. Because this design resulted in the least amount of anticipated issues between our three concepts generated, we decided to choose it as our method to secure the syringes to the holder.

Concept Description—Alpha Design

The in-phase slider mechanism is an exchange transfusion system utilizing a one-way valve configuration, a dual-syringe pumping mechanism, and one IV insertion point into the neonatal umbilical vein. One main objective of the in-phase slider ET device is to eliminate the necessity of having personnel present to manually perform syringe-pumping and valve-turning. Another main objective of the alpha design is to reduce the total number of syringe strokes for completion of a full exchange transfusion process. A third main objective of the alpha design is to have a system that eliminates the human factor when inserting blood into the patient by having a constant blood volume flow rate in and out of the patient, thus reducing both stress on the vein and blood cell lysis.

System Configuration

The configuration begins with an IV insertion point which consists of a 23-gauge needle into the umbilical vein connected to a 16-outer diameter gauge catheter within the umbilical cord. The catheter leads out of the umbilical cord to a three-way junction which splits the blood flow into two directions. In one direction is a waste-blood suction syringe and waste-blood disposal bag. In the other direction is a fresh-blood suction syringe and fresh-blood bag. Both waste-blood and fresh-blood sides of the system contain three-way junctions between the respective bag of blood, the respective suction syringe, and the initial three-way junction just outside of the umbilical cord. All junctions in the system are open to the flow in both directions. Equipped on two of the three sides of each junction are disposable one-way valves configured in a way to allow functioning of the system and to eliminate the need for manual toggling of the three-way junctions. Refer to Figure 12 for an embodiment of the configuration just described (p.28).

Process Description

The ET process is prefaced by priming the entire system with the blood intended to be transferred. The tubing, junctions, and valves need to be filled with blood before the process begins. The suction syringes also begin in their fully compressed position and operate simultaneously and in-phase with each other. Both syringe heads are connected to a slider mechanism which operates in the same direction as the syringe plunger and is powered by a CAM mechanism with a gear/crank and arm. Attaching the syringe heads to the same slider allows the syringes to operate simultaneously and in-phase with each other.

The cycle begins when the slider, and thus the syringe plungers, move back and the syringes intake fluid into their bodies. On the side with the fresh blood bag, fresh-blood is removed from the bag past the one-way valve and into the syringe. During this phase, blood is not removed from the neonate with this syringe because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood is removed from the neonate past the initial three-way junction, and into the syringe. During this phase blood is not removed from the waste-blood disposal bag because of the one-way valve configuration. This phase continues until the syringe plungers are fully extended back and the syringes are full with their respective fresh-blood and waste-blood.

The next phase of the cycle begins when the slider, and thus the syringe plungers move forward and the syringes force fluid out of their bodies. On the side with the fresh-blood bag, fresh-blood in the syringe is forced out past both three way junctions and into the neonate. During this phase fresh-blood is not forced back into the fresh-blood bag because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood in the syringe is forced out into the waste-blood disposal bag. During this phase blood waste-blood is not forced back into the neonate because of the one way valve configuration. This phase continues until the syringe plungers are compressed all the way and the syringes are empty of all their respective fluids. When this phase completes, the first phase begins again, and the cycle repeats.

Driving Mechanism

The driving mechanism of the slider is powered by a stepper motor which is energized and controlled by a microcontroller. A stepper motor was chosen over a servomotor because our system does not need a high speed motor to drive a 5 mL/min blood flow rate. The torque that the slider will require can be sufficiently provided by a stepper motor. Stepper motors are also used for their accuracy in low torque systems. Given the low amount of torque that the slider will ultimately need to drive the syringes, a stepper motor will be sufficient. The motor shaft will provide torque to a crank and a connecting rod will transmit the rotational torque into a translational force to move the slider. The microcontroller will have a user input in terms of blood volume flow rate output by the syringe, thus allowing flow rate to be controlled and held constant at a level suitable for the patient.
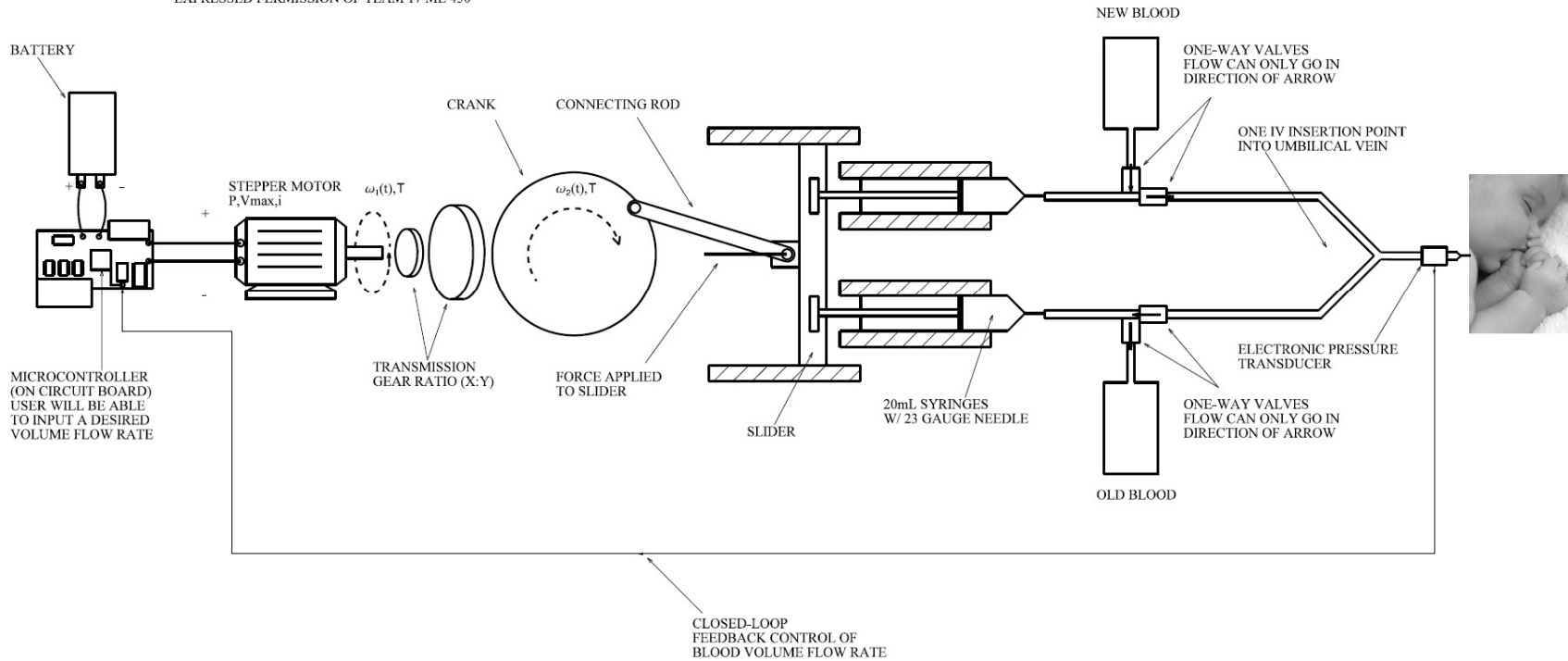
Figure 12. CAD drawing of alpha design (in-phase slider).

Blood Volume Flow Rate Control

A negative feedback, closed-loop control is included in the alpha design. The output blood volume flow rate will likely be less than the user inputted value due to energy losses between the stepper motor and the syringes. An electronic pressure transducer will read the pressure at the junction just outside of the umbilical vein and feed the signal back to the microcontroller. The pressure reading will be converted to a blood volume flow rate by the microcontroller using the tubing geometry. Then a proportional control will increase (or decrease) the power to the stepper motor, thus closing the gap between the inputted and outputted blood volume flow rates. (A proportional-derivative control and a proportional-integral-derivative control are being investigated as well.) The feedback control will also allow for compatibility with different syringe and tubing sizes and eliminate the need for blood volume flow rate to be a function of geometry. The feedback system will allow an automatic shutoff process to be implemented into the system as well.

Engineering Analysis

The following section will provide the vital analyses that must be completed to create a functional prototype.

Rigid Body Kinematics and Dynamics

The input to the pumping mechanism is ultimately the speed and torque of the motor. The volume flow rate is also directly proportional to the velocity of the slider. These two relationships can be interconnected using kinematics. A kinematics model will provide a relationship between the motor speed and the flow rate (i.e. the speed of the slider), which is essential when developing a feedback system based on the volume flow rate. The schematics of the motor would also be necessary to analyze the kinematic model. The relationship between the motor speed and the flow rate will be needed to determine the force analysis on this system.

For our design, it would be preferred that the power consumption is minimal; therefore the torque-angular velocity of the motor must be minimized as well. This would reduce the amount of power needed to be supplied by the motor, which would reduce the cost of the entire system.

Fluid Dynamics

Determining a model for the fluid flow is also a key element in our design. Since the flow rate is the key quantity that needs to be controlled, this should be the key element that should be analyzed. To analyze fluid flow within the tubing of the alpha design, a fluid flow model can be incorporated. A fluid flow model relates the fluid properties (i.e. density, viscosity) and the flow rate of the fluid. Before such an analysis can be completed, it is essential to determine whether the flow within the tubing is laminar or turbulent. Reynolds number (Re) can be utilized to determine the type of flow. Re is determined in equation 1 below, where ρ is the density of the fluid, V is the mean fluid velocity, D is the diameter in which the fluid flows, and μ is the dynamic viscosity of the fluid.

$$Re = \frac{\rho V D}{\mu} \text{ (Eq. 1)}$$

From Reynolds number, we determined that there would be laminar flow. Since the blood flow is laminar, Poiseuille's Law (Eq. 2) can be utilized to determine a relationship between fluid flow rate (Q) and the fluid properties,

$$Q = \frac{\Delta P \pi D^4}{128 \mu L} \text{ (Eq. 2)}$$

where ΔP is the pressure drop across a tube of length L and diameter D.

Modeling vs. Testing

Our first modeling endeavor will be a virtual simulation software such as *Adams* developed by MSC. This will give us a force analysis and will be able to tell us whether our forces are reasonable.

After the prototype is built, testing between the voltage of the motor and the output flow rate of the fluid will be conducted. This will be our most critical relationship. Since our actual motor will be a stepper motor rather than a servo motor, a comparison of the model with the actual system will be paramount. We will compare these results with our transfer function and make necessary changes.

Our next test will be to test the controller by manually changing the voltage of the motor and testing to see if the controller will correct the disturbance.

Design Drivers

Absolute design drivers consist of the motor and its transfer function. A motor that can actually drive the system is essential for it is what makes the system "automated." The motor must be able to function with the transfer function given by the feedback system. The transfer function is ultimately keeping the flow rate at the required value; therefore, it is vital to insure that the modeling is correct. Every other design requirements such as dimensions, power supply, and material will be determined off of this.

Parameter Analysis

The following sections will discuss the engineering analysis conducted on each component of our device to determine its potential success before manufacturing or purchasing took place. A thorough engineering analysis of the automated ET design was performed by evaluating basic engineering principles. (i.e. A stress analysis and a fracture analysis were performed on the components that would experience large loads.) The engineering analysis is quite basic because the device is relatively simple, and minimal forces and pressures are exerted within the system.

Mechanical Components

Linear Actuator: To measure the force required to move the syringe plunger in and out, we performed several tests in Professor Gillespie's lab using a force transducer. We used a vice to hold the syringe in place, and then pushed in the plunger when it was fully removed with the force transducer at a constant rate. We performed this experiment several times for each syringe, and tested approximately 5 different types and sizes of syringes. The maximum force required to push the most difficult plunger in was approximately 3 N. We used a factor of safety of 3 to account for additional frictional forces that would need to be overcome because of the slider, and also because removing the plunger was more difficult than pushing in the plunger. We were not able to measure the force required to remove the plunger because of equipment limitations (i.e. the force transducer could not perform this measurement). With this factor of safety, we concluded that we would need a driving mechanism capable of at least 10 N of force, or 2.25 lbs. of force.

Beyond measuring the plunger force of several syringes, we also measured the plunger length for each syringe using a pair of calipers. The biggest and longest syringe to be used with our device is a 25-mm syringe; therefore, we measured the plunger length of that syringe to determine the requirement of actuation for our driving mechanism. The plunger length for the 25-mm syringe was approximately 4 in, so we concluded that an actuation of at least 5 in. would be sufficient for our driving mechanism. However, we didn't want the driving mechanism to take up unnecessary space because minimizing the size of the device was one of our key customer requirements, so we also wanted to keep the actuation capability under 6 in.

After performing this parameter analysis on our syringes, we conducted online research to find the cheapest linear actuator we could find that met our minimum requirements. After extensive research, the best linear actuator for our system was found on www.surpluscenter.com. This linear actuator (Model Number LACT6P) has a 5.90 stroke length and a 107 lb. force capability. The stroke length was ideal for the actuation range we were hoping to find, and the 107 lb. force capability easily exceeds the minimum 2.25 lb. force required. This linear actuator was purchased for 79.95 USD, and satisfies all parameter analysis requirements for our driving mechanism. It also eliminates the need for a motor as one is built into the linear actuator already, thus it greatly helps with cost efficiency. The purchased linear actuator also contains a built in potentiometer, which will be utilized for measuring the volume flow rate of the system.

Track: A ball bearing track is utilized in the final design because a ball bearing track will prevent toggling between the slider and the track. Previously, a track consisting of two slots was designed, which is illustrated in Figure 13 (p. 32); however, after much analysis, the design had a great risk of failure due to toggling. We decided to use a ball bearing track made of steel, which is depicted in below (p. 32). A full analysis of the ball bearing track is provided in the sections below.

Friction: Friction is a major concern especially since the friction of the track will impact the blood volume flow rate. Since roller bearings are utilized in the track, the coefficient of rolling must be evaluated.
Assuming the worst case scenario where the frictional force is at a maximum, the maximum coefficient of friction can be determined. The maximum force due to friction will occur when the linear actuator is operating at its maximum force. The maximum force (F_{max}) utilized by the linear actuator is 10 N.

$$F_f = F_{max} - F_{needed} \text{ (Eq. 3)}$$

Equation 3 above expresses that the maximum frictional force (F_f) is equivalent to the force needed to move the slider (F_{needed}) subtracted from the maximum force utilized by the linear actuator (F_{max}). Assuming the standard friction relationship, then Eq. 4 below can be applied

$$F_f = \mu \, F_{normal} \text{ (Eq. 4)}$$

where μ is the coefficient of friction and F_{normal} is the normal force acting on the track. By combing Eq. 3 and Eq. 4 on can acquire the maximum coefficient of friction (μ_{max}) by means of Eq. 5 below.

$$\mu_{max} = \frac{(F_{max} - F_{needed})}{F_{normal}} \text{ (Eq. 5)}$$

For calculations specific to the final design, a maximum force of 10 N, a force needed of 3 N and the weight of the slider on top of the track being no more than 2 N provides a maximum allowable coefficient of friction to be 3.5.
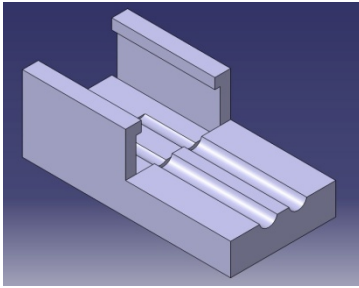
Figure 13. Previous track design, which was eliminated due to its high risk of toggling.
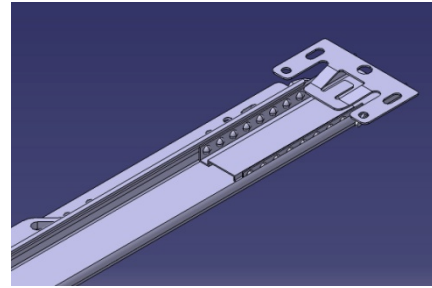
Figure 14. Ball bearing track design utilized in the final design.

The coefficient of friction for the rolling slider (μ_{roll}) is defined in Eq. 6 below

$$\mu_{roll} = \frac{RF}{r} \ \text{(Eq. 6)}$$

where *RF* is the rolling friction constant on the stainless steel and *r* is the radius of the rollers. *RF* for stainless steel is 0.039, while the *r* of the rollers is 0.40 mm. From this calculation, the μ_{roll} equals 0.0978, which is much less than the 3.5 maximum.

Toggling: Toggling is a major concern when dealing with sliding mechanisms. Toggle is due to high tolerances between moving parts, as well as high friction that could result in slipping.
The rolling slider is manufactured to have to have side walls on in all directions to ensure the slider will move only in the direction of the track. Through visual inspection the tolerances of the slider are far finer than those that could have been attempted from us in the shop; therefore, toggle will not be a problem with the track.

Slider: There are two modes of failure for the slider: fracture and yield due to bending. The fracture and bending of the slider will be evaluated.

Bending: The maximum bending moment would occur if there were a point load on the very end of the triangular plunger holder, which is depicted in Figure 15 below.
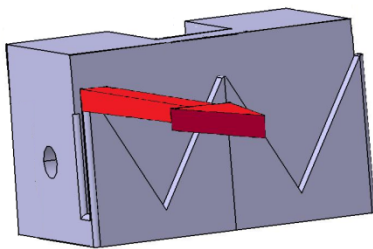
Figure 15. A 3D model illustrating the location of the maximum bending moment.
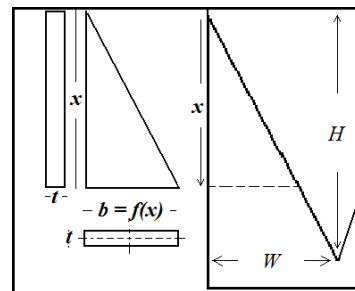
Figure 16. A layout of the slider's triangular cross-section.

To solve for the maximum bending moment at any arbitrary distance (*x*) from the point of the triangle as depicted in Figure 16 above. The base (*b*) of the cross sectional area becomes a linear function of *x* with the relation based on similar triangles depicted in Eq. 7 below.

$$b = f(x) = \frac{Wx}{H} \text{ (Eq. 7)}$$

The thickness (t) of the cross sectional area remains constant. The area moment of inertia (I) for any arbitrary cross-sectional area would have the following form (Eq. 8):

$$I = \frac{bh^3}{12} = \frac{Wxt^3}{12H} \text{ (Eq. 8)}$$

By taking a point load (P) at the tip of the triangle, the moment at any arbitrary point x down the triangle, would be equal to Px. The maximum bending stress (σ_{bs}) felt at each cross-sectional area would have the form following form:

$$\sigma = \frac{Mc}{I} \text{ (Eq. 9)}$$

where M is the moment and c is the distance from the center of the cross-sectional area to the top edges which is equivalent to half the thickness. At any arbitrary distance x down from the tip of the triangle, the maximum bending stress would be as depicted in Eq. 10 below.

$$\sigma_{bs}(x) = \frac{PH}{2Wt^2} \text{ (Eq. 10)}$$

To determine the maximum bending stress, a risk factor of 3 was multiplied into Eq. 10 above. From Eq. 10, it was determined that the σ_{bs} will equal 1.06 MPa; therefore, the yield strength material property must be greater than the maximum bending stress. The slider is being made from ABS, which has a yield strength of 41 MPa, which will exceed expectations.

Fracture mode: Another failure mode could be due to crack fracture. A basic, first approximation would be to simplify the triangle holder with a plane stress mode of the order equivalent to the 1.06 MPa maximum stress calculated in the aforementioned section.
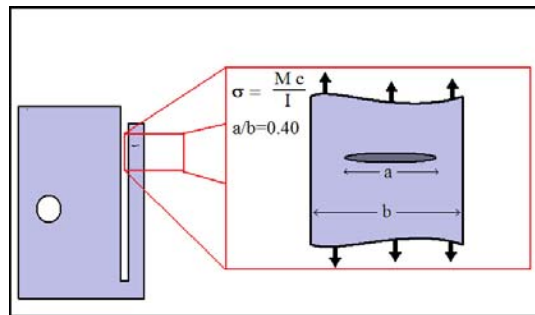
Figure 17. Worst possible crack scenario for the slider's triangle face

If the worst case scenario is assumed (depicted in Figure 17above), where an inherent crack already exists in the material that is 40% the width of the material then the resulting minimum fracture toughness required by the material can be determined using Eq. 11 below.

$$K_{min} = \sigma \cdot \sqrt{\pi \cdot 0.40 \cdot a} = 1.06 MPa \cdot \sqrt{\pi \cdot (0.40) \cdot 3.175 \cdot 10^{-3}m} = 0.0670 \, MPa\sqrt{m} \text{ (Eq. 11)}$$

The fracture toughness of ABS plastic is 3.23 MPa$\sqrt{m}$, which is much great than 0.0670 $MPa\sqrt{m}$; therefore, the design will not fail due to crack fracture. For Ghanaian production, the slider should be

manufactured out of PVC, a readily available material in Ghana. (Please Reference Appendix D for a thorough material analysis.)

Full Mechanical Sub-System: The mechanical portion of our design has to weigh less than a specified value as well as be smaller than a specified volume.

Weight: The material used to create the mechanical subsystems is limited based on the weight constraint. The mechanism must weigh less than 65 N. If the worst case scenario is assumed, where the entire ET mechanism is made of the same material, then the maximum density can be determined using the total volume of the system as well as the targeted weight requirement. The maximum density of the material will provide a density specification, which can be utilized to find a viable material. For the final design, Cambridge Engineering Software was used to determine a sufficient material.

The weight of the motor (13.2 N) and the track (2.2 N) are known; therefore, the maximum density can be determined using Eq. 12 below.

$$\rho_{max} = \frac{Weight_{Target}/R.F. - Track - Motor}{V_{total} \cdot g} \text{ (Eq. 12)}$$

where the *Weight$_{Target}$* is the target weight of the system, or 65 N in this case, *RF* is the safety factor of 1.2, *Track* is the weight of the track, *Motor* is the weight of the motor, *V$_{total}$* is the total volume of the material that makes up the system ($2.51 \cdot 10^{-3}$ m^3 for the housing and less than $0.23 \cdot 10^{-3}$ m^3 for all of the components for a total of $2.74 \cdot 10^{-3}$ m^3), and *g* is the gravitational acceleration (9.8 m/s^2). From Eq. 12, the maximum allowable density is 1442 kg/m^3.

Both materials being used in the prototype and final design, ABS plastic and PVC, fit material requirements. ABS plastic has a density of 1,135 kg/m^3 and PVC has a density of 1,400 kg/m^3 which are both less than 1442 kg/m^3. For Ghanaian production, the slider, syringe holder, and casing should be manufactured out of PVC, a readily available material in Ghana and the casing. (Please Reference Appendix D for a thorough material analysis.)

Electrical Components
There are various electrical components involved in the operating system of the automated ET device. The operating system will allow functionality based on various user inputs, such as the required blood volume flow rate, amount of blood of be exchanged, and the volume of the syringes. The operating system will also allow the system to function based on feedback; therefore, insuring that the ET device functions at the correct rates and requirements. The schematic of the electrical system is provided in Appendix E-1. The electrical system contains the following components: the microprocessor, the oscillator, the Universal Serial Bus (USB), the linear regulator, the LCD, the power plug, the LED, the buzzer, the mini header, the motor driver board, the microSD, the rotary encoders, the motor, and the potentiometer. From the schematic of the electrical system, a circuit board diagram was created by placing the electronic components correctly on the circuit board. The circuit board diagram is illustrated in Figure 18 below. The circuit board diagram was then outsourced to be made.
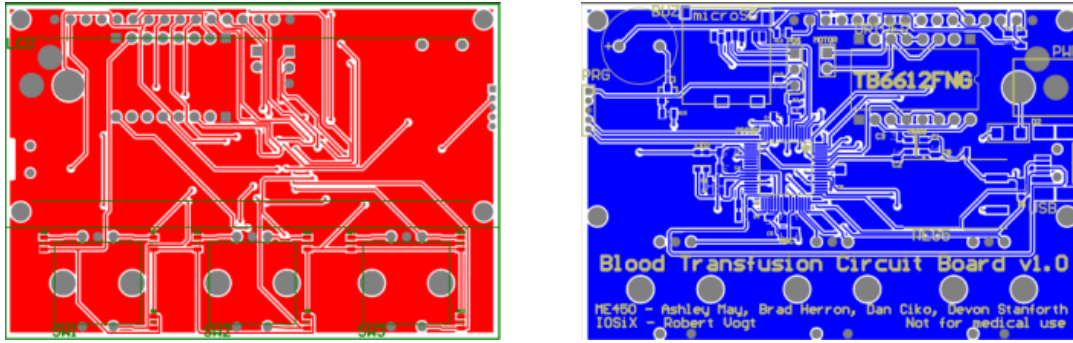
Figure 18. Front and back view of the circuit board diagram.

Microprocessor: The main component of the operating system is the microprocessor. The microprocessor is the central processing unit (CPU), which controls the logic of the device. The microprocessor consists of various interfaces. The control unit of the microprocessor functions based on a set of instructions, which orchestrates the operations of the other units. The execution path taken by the control unit can depend upon the status bits produced by the various input components. The microprocessor also serves as a memory interface, thus allowing programs and data to be stored. The microprocessor involves an interrupt or exception controller, which enables the microcontroller to respond to requests from the external environment or to error conditions by allowing interruption of the ongoing process. The interrupt/exception controller will detect system malfunction, as well as the end of cycle; therefore, the interrupt/exception controller will yield all operations and sound an alarm to inform the medical personnel. To insure the microprocessor functions correctly, instructions will be programmed into the microprocessor using C programming language. There are various types of microprocessors, which vary primarily based on the number of bits the processor operates on at one time. Cost and the number of bits are the two main concerns in determining the optimal microprocessor for the automated ET device design. Based on these two concerns, the PIC32MX440F512H microprocessor was determined to be the optimal microprocessor for our design. It is a 32-bit flash microcontroller. (The specifications of the micrprocessor are provided in Appendix E-2.)

Oscillator: The electronic oscillator acts as a frequency control. The electronic oscillator provides an output frequency of 4 MHz to the microprocessor, which amplifies the frequency of the microprocessor. The amplification of the microprocessor frequency acts as an electronic filter, which filters out noise travel created by the electromagnetic devices. A schematic of the electronic oscillator is provided in Figure 19 on page 35. The oscillator (part number SIT8002AI-23-33E-4.00000T) was chosen based on its frequency output, as well as its ability to operate with 3.3 V. (The specifications of the oscillator are depicted in Appendix E-3.)
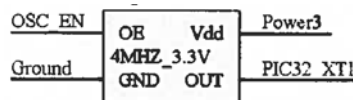


Figure 19. Electronic schematic of the oscillator.

Universal Serial Bus (USB) Connector: The Universal Serial Bus (USB) connectors enables the microprocessor to be connected to a host computer; therefore, allowing the microprocessor to be reprogrammed. The schematic of the USB is depicted in Figure 20 below. The USB is connected to the microprocessor through the "USB D+" and "USB D-" connectors, which is depicted in Fig. # (p.#). USBs also provide power to low-consumption devices; therefore, the USB outputs 5V into the circuit, which is depicted in Figure 21 below. The 5V output is connected to the linear regulator. The USB is also

35

grounded. USB connectors are fairly standards; therefore, a USB connector with five circuits, with a SMT solder tail, and of minimal size was chosen. The specifications of Model 538-67503-1020 USB connectors is provided in Appendix E-4.
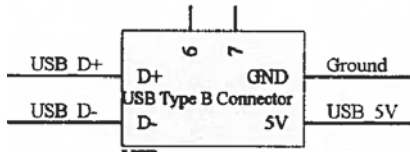

Figure 20. Electronic schematic of USB connector.
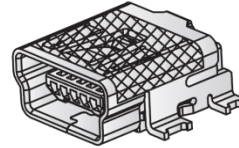
Figure 21. USB connector used in circuitry.

Linear Regulator: The linear regulator acts as a voltage regulator. The linear regulator is made to act like a variable resistor, continuously adjusting a voltage divider network to maintain a constant output voltage. The linear regulator used is a shunt regulator, which operates by way of the Schottky diode's action of maintaining a constant voltage across itself when the current through it is sufficient enough. The Schottky diode is a semiconductor diode with a low forward voltage drop and a very fast switching action. A normal diode has between a 0.7 and a 1.7 voltage drop, while a Schottky diode voltage drop is between approximately 0.15 and 0.45—this lower voltage drop translates into a higher system efficiency. The voltage output of 5V from the USB connector is input into the linear regulator, where it is then converted to 3.3V before being outputted into the microprocessor. (The schematic of the linear regulator is depicted in Figure 22 on page 36.)
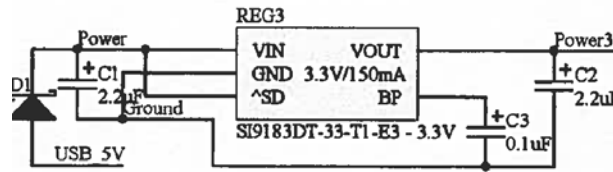
Figure 22. Electronic schematic of linear regulator.

The Schottky diode was chosen primarily due to its high efficiency. Schottky diode part number MSS1P4-E3/89A is used in the circuitry of the automated ET device. (The specifications of the Schottky diode are depicted in Appendix E-5.) The linear regulator used for the automated ET device is part number SI9183DT-33-T1-E3-3.3V, which provides a maximum voltage output of 3.3 V, a maximum current output of 150 mA, and a dropout voltage of 135 mV. This linear regulator is ideal because it outputs 3.3 V, which is the voltage required by the microprocessor. (The specifications of the linear regulator are illustrated in Appendix E-6.)

Liquid Crystal Display (LCD): A liquid crystal display (LCD) is an electronically-modulated optical device shaped into a thin, flat panel made up any number of color or monochrome pixels filled with liquid crystals and arrayed in front of a light source or reflector. An LCD will be utilized to display various values, such as the volume flow rate and the amount of blood transfused. A 16 character by 2 line display with a blue background and a yellow backlight was chosen to display important data. LCD part number GDM1602K was chosen primarily due to its low cost and its capability of showing 32 characters on its screen. The LCD is illustrated in Figure 23 below. The electronic schematic of the LCD is depicted in Figure 24 below. All of the inputs and outputs of the LCD are connected to the microprocessor, except for

the 5V power, which is an output of the voltage regulator. (The specifications of the LCD are provided in Appendix E-7.)
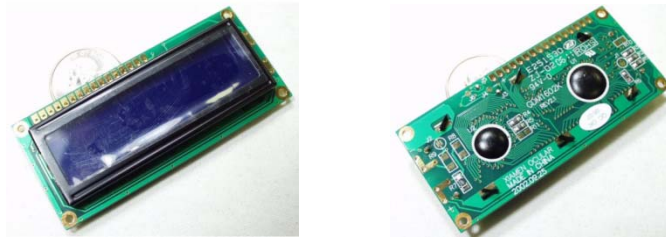

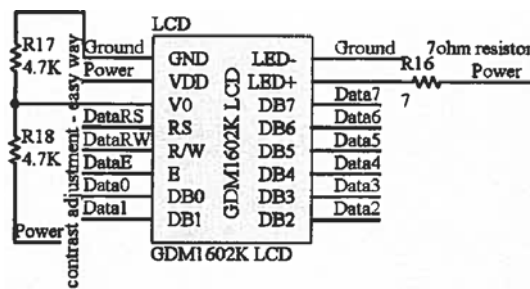**Figure 23. Front and back view of the LCD.**

Figure 24. Electronic schematic of the LCD.

Power plug: The automated ET device will be powered by AC power; therefore, the AC power must be converted to DC power. The power plug encompasses the following two components: the AC adapter and the DC power jack. The AC adapter, depicted in Figure 25 (p. 37), will convert the input voltage of 120 VAC to 12 V DC. The AC adapter is then connected to the DC power jack, depicted in Figure 26 (p. 37), which transfers the 12V DC to the circuitry. For this device to be compatible with Ghanaian power outlets, an AC adapter will need to be of the correct orientation. The AC Adapter will also need to convert the 220 V, 50 Hz AC power to 12 V DC power. For Ghanaian use, the AC Adapter (Part number EMS120050-P5P-SZ) should be purchased. This AC Adapter will be compatible with the DC power jack as well.
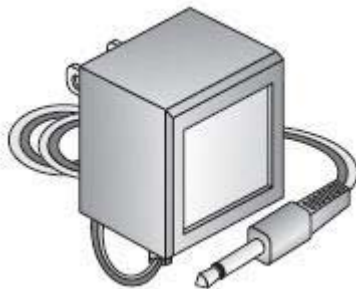
Figure 25. AC adapter used to convert AC current to DC current.

Figure 26. DC power jack used to convert current.

The microprocessor cannot handle 12 V; therefore, a rectified diode and a voltage regulator are implemented into the circuitry to decrease the voltage output. A rectifier diode is employed to convert AC current to DC current. The rectifier diode part number NRD4001 was chosen primarily based on cost and functionality. (The specifications of the rectifier diode are provided in Appendix E-8.) The voltage regulator serves the purpose of automatically maintaining a constant voltage level. The voltage regulator must output a voltage of 5V; therefore, voltage regulator part number L7805CD2T was chosen for this application. Voltage regulator part number L7805CD2T outputs a current of 1.5A and a voltage of 5V. This regulator also provides short circuit protection, as well as thermal overload protection. (The specifications of the voltage regulator are provided in Appendix E-9.) The voltage regulator then outputs 5V to the LCD. The schematic of the power plug and voltage regulator is provided in Figure 27 below. Part number PJ-202A was utilized as the DC power jack. (The specifications of the DC power jack are provided in Appendix E-10.)
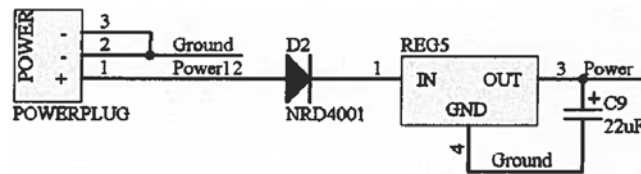
Figure 27. Electronic schematic of power plug.

Light Emitting Diode (LED): A light emitting diode (LED) is an electronic light source. The LED will be used to inform the user whether the device is on, off, or has completed its cycle. In order to indicate the three settings of the system, a two color LED will be required. The chosen LED (part number CMD15-22SRUG) emits the colors red and green through a clear lense; therefore, the LED will not emit any light when the power is off, the LED will be programmed to emit green when the device is running, and lastly, the LED will be programmed to emit high energy red when the device has completed its cycle. The LED will be connected to the microprocessor through the following two main connections, which are depicted in Figure 28 (p. 38): 100 Soft LED and 100 Soft LED2. The microprocessor will supply 3.3V to operate the LED. (The specifications of the LED are provided in Appendix E-11.)
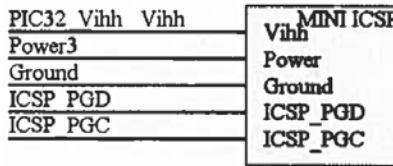

Figure 28. Electronic schematic of the LED.

Buzzer: The buzzer will sound at the end of the process to notify medical personnel. The buzzer chosen for this application utilizes a transistor. The buzzer sounds based on the actuation of the transistor by a frequency of 2 kHz. The continuous connection and disconnection of the transistor creates a vibration effect, which produces approximately a 95 dB sound. The transistor connects and disconnects by means of the "BUZSIG" connection illustrated in Figure 29 below. (The specifications of the transistor are provided in Appendix E-12.) A magnetic buzzer (part number CEM-1203) was utilized for the automated ET device because it could be easily implemented into the circuitry. (The specifications of the magnetic buzzer are provided in Appendix E-13.)
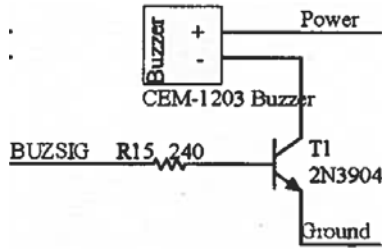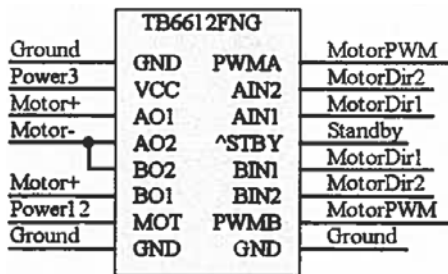
Figure 29. Electronic schematic of the buzzer.

Mini Header: The mini header provides the source code used to program the microprocessor. Although the microprocessor can be reprogrammed through the use of the USB connector or the card connector microSD, the mini header provides the initial code used to program the microprocessor. The mini header operates from a voltage source of 3.3 V. The mini header then outputs the source code to the microprocessor, which is illustrated in Figure 30 below. The mini header (part number 2211S-08G) was chosen based on its compatibility with the microprocessor. (The specifications of the mini header are provided in Appendix E-14.)

Figure 30. Electronic schematic of the mini header.

Motor Driver: The motor driver controls the motor by means of starting and stopping the motor, selecting forward or reverse rotation of the motor, selecting and regulating the speed of the motor, regulating or limiting the torque of the motor, and protecting the motor from overloads and faults. The motor driver is connected to the motor through the "Motor +" and "Motor -" connections depicted in Figure 31 (p. 39). The motor driver operates with a maximum power voltage of 15 V; therefore, both the 12 V DC source and the 3.3 V source are utilized in powering the motor driver. The motor driver is then connected to the microprocessor through the "MotorPWM," "MotorDir2," and "MotorDir1" connections illustrated in Figure 31 below.

**Figure 31. Electronic schematic of the motor driver.**

**Figure 32. Motor driver used to control the DC motor.**

The microprocessor is responsible for instructing the motor driver, in which the motor driver then alters the motor parameters as indicated by the microprocessor. The motor driver (part number TB6612FNG), illustrated in Figure 32 above, was chosen based on its voltage operating range of 4.5 to 15.0 V, its peak

output current of 3.2 A, and its cost. (The specifications of the motor driver are provided in Appendix E-15.)

Card Connector MicroSD: A card connection microSD (illustrated in Figure 33 below) will be utilized much like that of the USB connector. The card connector microSD acts as a backup method for reprogramming the microprocessor. The card connector microSD is powered by 3.3 V as depicted in Figure 34 below. The card connector is then connected to the microprocessor through the following connections as illustrated in Figure 34 below. : "SPI_DO_3," "SPI_CLK_3," "SPI_DI_3," and "SPI_CS_3." These four connections are utilized to reprogram the microprocessor. The card connector microSD (part number 2908-05WB-MG) primarily based on its cost. (The specifications of the card connector microSD are provided in Appendix E-16.)
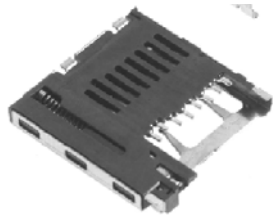

**Figure 33. Card connection microSD used in circuitry.**
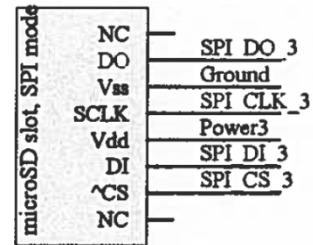
Figure 34. Electronic schematic of card connector microSD.

Rotary Encoders: A rotary encoder is an electro-mechanical device used to convert the angular position of a shaft to an analog or digital code. Three rotary encoders are used for user input. The rotary encoders will allow the user to turn the device on and off, as well as input the optimal blood flow rate, the total volume of blood to be exchanged, and the minimum and maximum positions the linear actuator will need to move to insure that exactly one full stroke is achieved. The rotary encoders output data to the microprocessor through the "ROTA," "BTN," and "ROTB" connections illustrated in Figure 35 below. Rotary Encode (part number RE130F-41-175F-12P), illustrated in Figure 36 below, was chosen primarily based on cost and size. (The specifications of the rotary encoders are provided in Appendix E-17.)
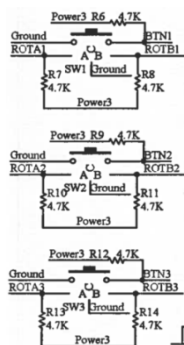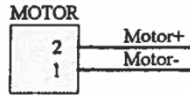

Figure 35. Electronic schematic of the rotary encoders.
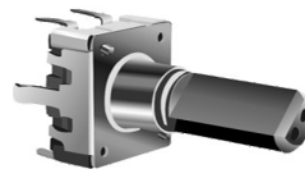

Figure 36. Rotary encoder used by medical personnel to set ET parameters.

Motor: The DC motor is provided with the linear actuator. The DC motor has two connection points in which the motor driver is connected. The schematic of the motor is provided in Figure 37 on page 41.

Figure 37. Electronic schematic of the motor.

Potentiometer: The potentiometer is also provided within the linear actuator. The potentiometer serves great importance within the automated ET device because it determines the distance the plunger has travelled within a given amount of time. The microprocessor can then take the given distance and time, and calculate the volume flow rate. Based on this volume flow rate, the microprocessor can then instruct the motor driver to either slow down or speed up the motor speed. A schematic of the potentiometer is provided in Figure 38 below.
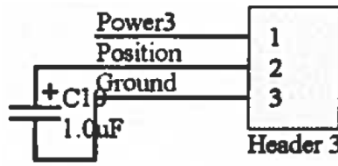
Figure 38. Electronic schematic of the potentiometer.

Medical Components

A majority of the medical supplies utilized by the automated ET device will be provided by KATH; however, the medical supplies offered at KATH are unavailable for the purpose of experimentation and analysis. Medical supplies similar to those provided by KATH must be employed to determine functionality of the automated ET device. The following medical supplies are available at KATH: syringes, tubing, catheters, needles, and IV bags. These medical supplies are fairly standard; therefore, the functionality of the automated ET device should not be affected by the use of slightly different medical supplies than those provided at KATH. For the automated ET device to function properly, the following two medical components must also be implemented at KATH: one-way valves and Y-connectors.

Syringes: Syringes vary geometrically and dimensionally from manufacturer to manufacturer. Syringes of the same volume are typically very similar in size and shape; however, there often slight discrepancies between syringes coming from different suppliers. Due to the inconsistencies between manufacturers, the automated ET device will accommodate all 20-25 mL syringes. The prototype will be tested with a variety of 20 and 25 mL syringes to insure that the syringes at KATH will be compatible with the automated ET device.

Tubing: KATH presently uses tubing with an inner diameter (ID) of 2.5 ± 0.5 mm when performing a neonatal ET. To insure that the tubing at KATH will be compatible with the automated ET device, tubing with an inner diameter of 2.5 mm will be utilized in the system. The 2.5 mm ID tubing was donated by DirectMed.

Catheters: Assistant Professor Kathleen Sienko acquired a catheter from KATH during her visit to Ghana in February 2009; therefore, the acquired catheter will be implemented into the automated ET system. If the catheter fails, Robin Jahnke provided us with a couple catheters from the neonatal ET package, which can be implemented if needed. The catheter should not affect the volume flow rate of the blood; therefore, the slight discrepancies between the catheter used for the prototype and the catheter used at KATH should have a negligible effect.

Needles: KATH presently uses 21 or 23 gauge size needles when performing a neonatal ET; therefore, the automated ET device prototype must also be tested with 21 and 23 gauge size needles. Assistant Professor Sienko also acquired needles from KATH during her visit to Ghana; therefore, these needles will be utilized in the automated ET system. 21 and 23 gauge needles are standard sizes, which will make it easy to acquire more needles if needed.

IV bags: IV bags are presently used at KATH to dispose of the waste blood; therefore, an IV bag will be needed for the automated ET system. An IV bag does not affect the performance of the exchange transfusion. The IV bag must be large enough to accommodate all of the waste blood and it must connect to the system properly to insure that the blood does not leak from the system. 2 Liter Graduated IV Bags (Model Number BA-006) IV bags are being provided by DirectMed, which will be utilized in the automated ET system.

One-way (check) valves: One-way valves must be implemented into the automated ET system to prevent the fresh blood from mixing with the waste blood. One-way valves are not currently available at KATH; however, one-way valves must be implemented to insure the proper function of the automated ET device. Due to the extremely small tubing being used, the one-way valves must mate with the tubing. The one-way valves utilized by the automated ET device have a 3.175 mm diameter; therefore, the one-way valves will act as the female component (surrounding the tubing—the male component). The tubing and the one-way valves should mate properly because the tubing has an inner diameter of 2.5 mm; therefore, the outer diameter of the tubing will exceed 2.5 mm coming very close to 3.175 mm. The one-way valves (Model Number AG090312, Item Number 64046) are being supplied by the United States Plastic Corporation.

Y-connectors: Y-connectors are required to insure the functionality of the automated ET device. The Y-connector provides a junction between the fresh blood and waste blood insert, which insures that the neonate only needs one IV insertion point. The Y-connectors will also be used to connect the blood bags to the automated ET device. The Y-connectors will have an inner diameter of 2.5 mm to insure a tight connection between the Y-connector and the 2.5 mm ID tubing. The Y-connector will be a male component; therefore, it will insert into the tubing. Y-connectors (Model Number CY-001) are being supplied by DirectMed at no cost. An engineering drawing of the Y-connector is provided in Figure 39 below.
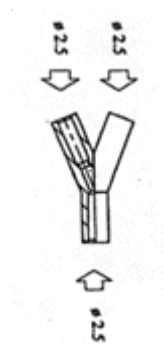
Figure 39. Model CY-001 Y-connector provided by DirectMed.

Final Design Description

Our design consists of both mechanical and electrical subsystems. The purpose of the mechanical subsystem is to actuate the syringe plungers and cause the flow of blood to and from the patient. A secondary mechanical subsystem is that of the blood pathway between the syringes, blood bags, and patient. The blood pathway subsystem is a unique configuration of tubing, one-way valves, and a Y-

junction connectors. The purpose of the electrical subsystem is to power and control the linear actuator and its function of driving the mechanical sliding mechanism. The result of the three subsystems is a mechatronic system that completes the overlying objective of exchanging the blood of a neonatal patient. A diagram of the entire system is depicted in Figure 44 on p.46.

System Configuration

The configuration begins with an IV insertion point which consists of a 23-gauge needle into the umbilical vein connected to a 16-outer diameter gauge catheter within the umbilical cord. The catheter leads out of the umbilical cord to a three-way junction which splits the blood flow into two directions. In one direction is a waste-blood suction syringe and waste-blood disposal bag. In the other direction is a fresh-blood suction syringe and fresh-blood bag. Both waste-blood and fresh-blood sides of the system contain three-way junctions between the respective bag of blood, the respective suction syringe, and the initial three-way junction just outside of the umbilical cord. All junctions in the system are open to the flow in both directions. Equipped on two of the three sides of each junction are disposable one-way valves configured in a way to allow functioning of the system and to eliminate the need for manual toggling of the three-way junctions. Refer to Figure 44 for an embodiment of the configuration just described (p.46).

Process Description

The ET process is prefaced by priming the entire system with the blood intended to be transferred. The tubing, junctions, and valves need to be filled with blood before the process begins. The suction syringes also begin in their fully compressed position and operate simultaneously and in-phase with each other. Both syringe heads are connected to a slider mechanism which operates in the same direction as the syringe plunger and is powered by a linear actuator. Attaching the syringe heads to the same slider allows the syringes to operate simultaneously and in-phase with each other.

The cycle begins when the slider, and thus the syringe plungers, move back and the syringes intake fluid into their bodies. On the side with the fresh blood bag, fresh-blood is removed from the bag past the one-way valve and into the syringe. During this phase, blood is not removed from the neonate with this syringe because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood is removed from the neonate past the initial three-way junction, and into the syringe. During this phase blood is not removed from the waste-blood disposal bag because of the one-way valve configuration. This phase continues until the syringe plungers are fully extended back and the syringes are full with their respective fresh-blood and waste-blood.

The next phase of the cycle begins when the slider, and thus the syringe plungers move forward and the syringes force fluid out of their bodies. On the side with the fresh-blood bag, fresh-blood in the syringe is forced out past both three way junctions and into the neonate. During this phase fresh-blood is not forced back into the fresh-blood bag because of the one-way valve configuration. On the side with the waste-blood bag, waste-blood in the syringe is forced out into the waste-blood disposal bag. During this phase blood waste-blood is not forced back into the neonate because of the one way valve configuration. This phase continues until the syringe plungers are compressed all the way and the syringes are empty of all their respective fluids. When this phase completes, the first phase begins again, and the cycle repeats.

With the general configuration of the system and the exchange transfusion process described, the components that make up the entire system will now be described. To aid in understanding all of the components in the system and their interactions with one another, the system will be divided up in mechanical, electrical, and tubing subsystems. In the mechanical subsystem section, the individual components will be described, and their interactions with each other will be described as well. In the electrical/circuit subsystem section, the aspects that directly resolve engineering specifications will be

described, while the other electrical support components are described in the preceding *Parameter Analysis of Electrical Components* section. In the tubing subsystem section, a brief evaluation has been given, however reference to preceding sections is given as adequate description for final design.

Mechanical Subsystem

The mechanical subsystem consists of a holding block, a slider, a track, and a linear actuator. The subsystem can be seen in Figure 40 (p. 44) with two 25 mL syringes in place. Not included in the picture is the linear actuator, which is also a part of the mechanical subsystem, and is connected to the slider with a pin. The linear actuator is depicted in Figure 41(p. 44). The mechanical subsystem is the only system out of the three that contains machined parts by the ME 450 exchange transfusion team. The drawings of all fabricated parts are properly located in the Appendix F.
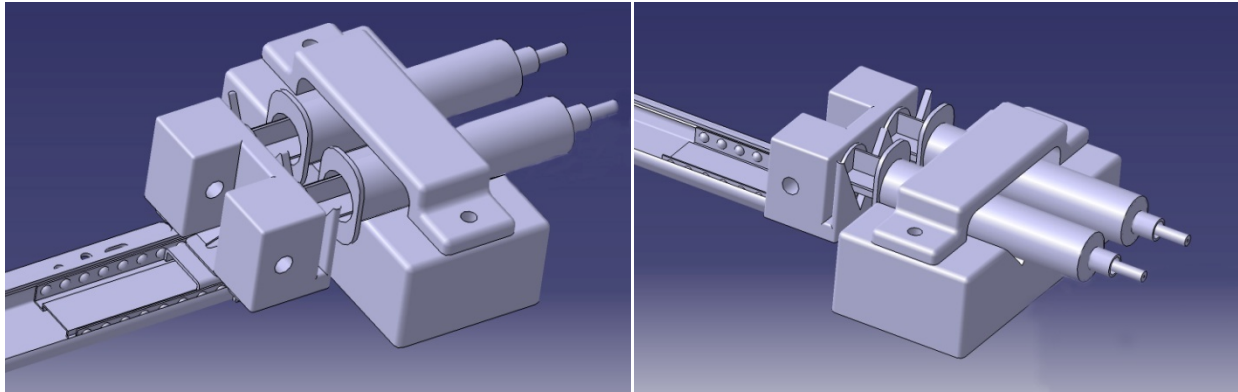
Figure 40. Assembled mechanical subsystem, minus the linear actuator.

Linear Actuator: The linear actuator is a considerable step forward from the crank and connecting rod driving mechanism of the alpha-design. The linear actuator is designed specifically to accommodate the motions needed for our design, without all of the kinematics and power transmission in the middle. The linear actuator will be connected to the slider by a pin and the velocity of the shaft will be micro-controlled to obtain the desired volume flow rate of the syringe (will depend on the diameter of the syringe). The motor itself is 12 V DC and at the loads expected for our application the shaft can actuator

Figure 41. 5.90" Stroke 107 lb. 12 V DC Linear Actuator

up to 0.59 in/sec. For the maximum desired flow rate and for a 25 mL syringe diameter, a linear velocity of 0.00088 in/sec is needed. This posed a lower limit, rather than an upper limit problem with the motor speed. However, after analysis of the voltage resolution out of the micro-driver, we calculated that the motor will be able to handle speed low enough for our application.

The linear actuator comes equipped with a position potentiometer as well, which will come into play with our feedback control schedule (described in section title *Electrical/Circuit Subsystem*).

Lastly, the actuator is attached to the main board with a light duty mounting bracket, which can be seen below in Figure 42 and is referenced in our BOM (Appendix B).

Figure 42. Commercial mounting bracket supplied by same manufacturer of linear actuator.

Slider and Track: The slider is the component that the syringe plungers attach to. It also is the intermediate object that transmits the force of the actuator to the plungers. It can be seen in Figure 43 above, and is dimensioned in Appendix F. The important feature of the slider is its ability to accept and attach to it various size plungers, while not using any loose parts to do so (which can be lost during the process). The solution to this is the V-shaped slot design which can be seen on the front face of the slider. Figure 45 shows how the plunger is supposed to attach into the slot (p. 47).
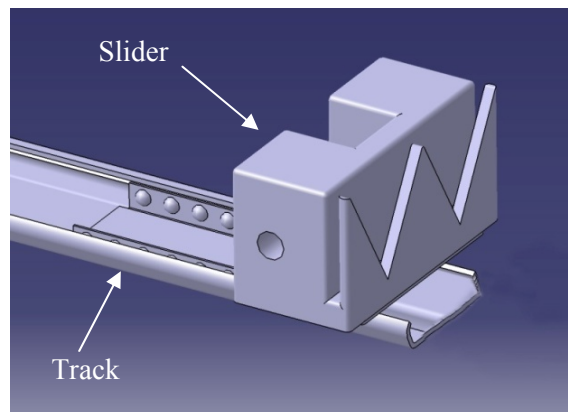
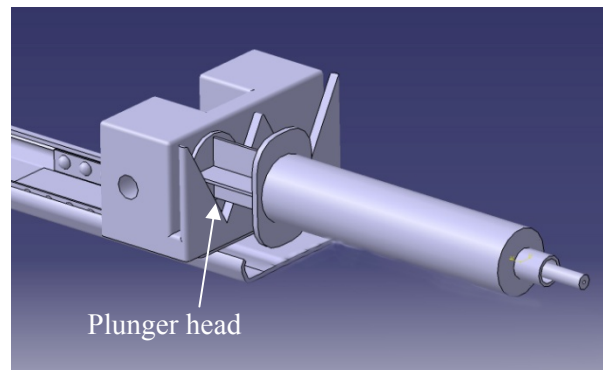**Figure 43.** Slider and track components in contact with each other.
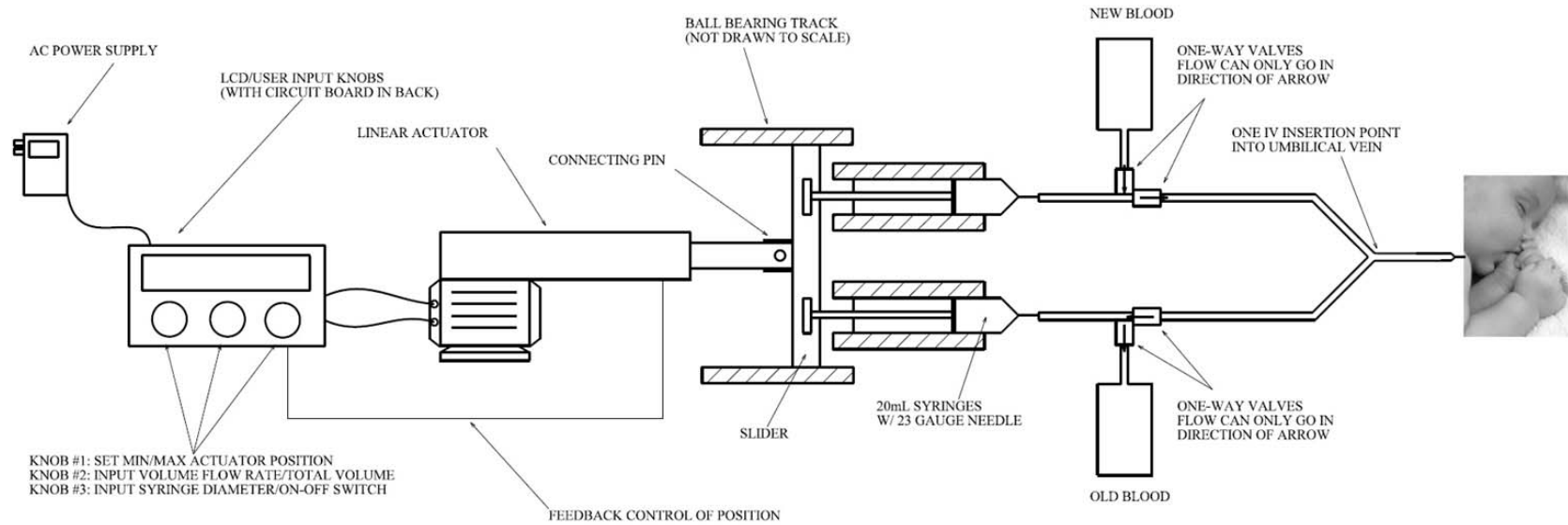
Figure 44. System diagram of the final design.

Plungers of different diameters will simply fall into the slot until the V design stops it from going any further. This method, as opposed to a uniform slot will make the slider compatible with a much larger variety of plungers.


Figure 45. Plunger heads of various size can fit into and remain in the V-shaped slot.

The track component depicted Figure 46 below complements the slider and provides a means for it to move smoothly and as effortlessly as possible. The track is a 3/4" extension slide with lift-out release, which can be purchased (see BOM) and can be cut down to size per the initial fabrication section. The slider attaches to the platform on the track with four screws and the platform itself rides on a series of low friction ball bearings. This setup is shown in Figure 47 (p. 48).
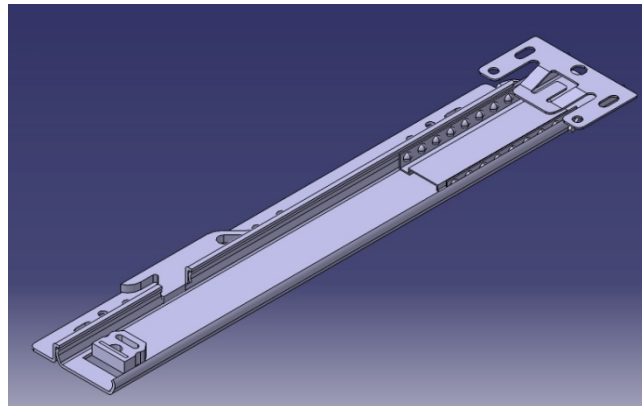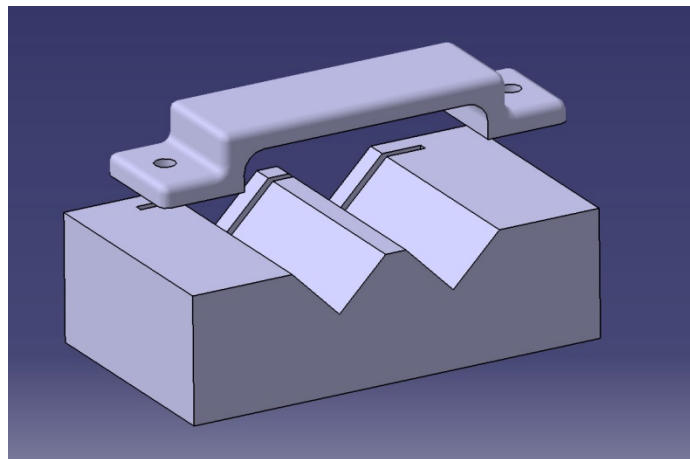

Figure 46. 3/4" extension cabinet slide with lift-out release from McMaster-Carr (see BOM).
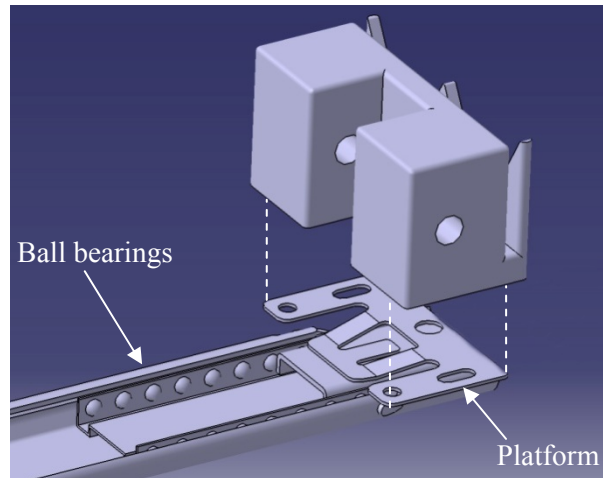
Figure 47. The slider attached directly the platform that slides on the track.

Holding Block: The holding block's purpose is to hold the syringes in place while the plungers are actuated by the slider. The holding block with brace is shown below in Figure 48.


Figure 48. Holding block with brace.

The holding block has a similar groove design to the slider. It utilizes V-shaped grooves to allow the placement of various diameter syringes. After two identical syringes (20-25 mL recommended) are placed in the grooves and their collars are placed within the holding slot (Figure 49, p. 49), the brace is brought down onto the syringes and fastened (Figure 50, p. 49). It should be noted that the bracer is meant only to keep syringes from escaping in the vertical direction. As long at the bracket is snug against the syringes, the brace has completed its task, even if the legs of the brace don't come in contact with the holding block.
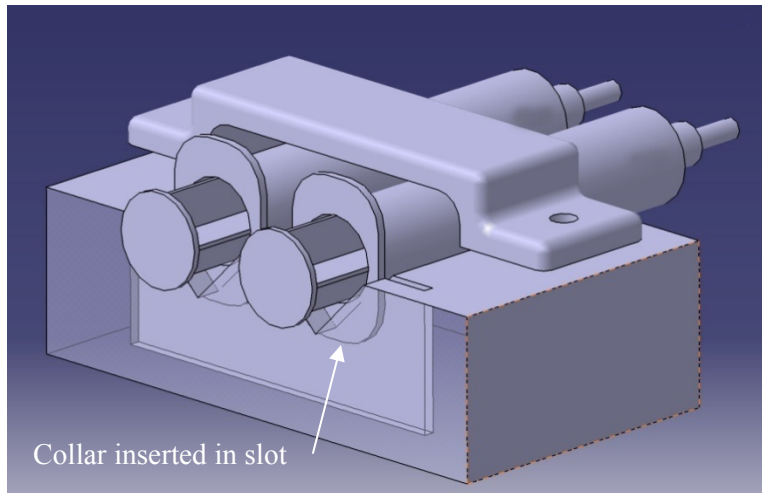
Figure 49. Syringes in place in the holding block, with collars contained in the slot, and syringe bodies held down by brace.
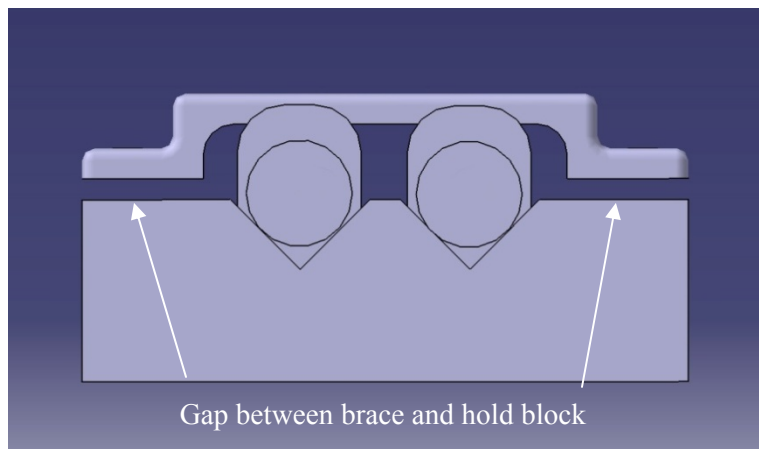
Figure 50. Brace is flexible and need not be tightened all the way to the holding block.

Casing (Enclosure) and Base Board: The base board is the plane that the linear actuator, track, and syringe holder are fastened to. The board is necessary to keep the four components stationary and relative to each other (see manufacturing and fabrication place for precise dimensions). The casing will be a box with dimensions 23.00" x 6.00" x 4.00" and has the circuit board and LCD placed within its sides for ease of use as well as an opening in the front for the tubing and a hole in the back for a power cord. The walls are made up of box joints and fastened to the base. For demonstration purposes, our prototype's walls are made up of plexi-glass to allow users to view the process, whereas the final design's walls would be made up of PVC, due to its higher durability (allowing greater protection for the components if assembly was ever dropped) as seen in Figure 51 below. The base board would also be made up of PVC for the final design due to the same reason; however the prototype has a base board made of wood to allow alterations to be made easier if needed.
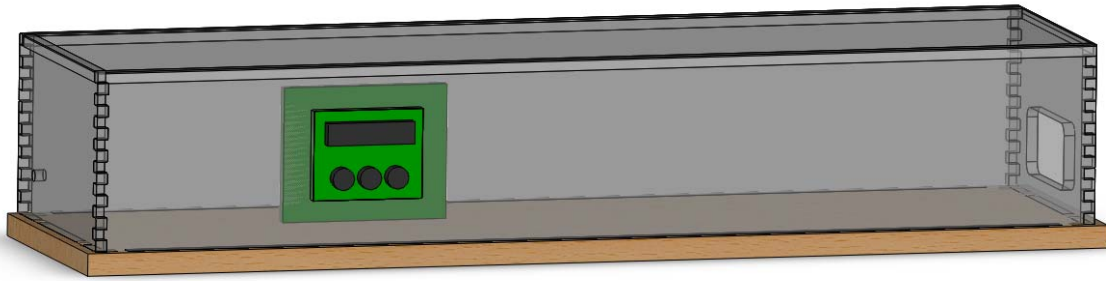
Figure 51. View of prototype's container with LCD and computer chip.

Electrical/Circuit Subsystem

While the mechanical subsystem is the bones of the exchange transfusion system, the electrical/circuit subsystem is the lifeblood. An electrical aspect to an automated exchange transfusion device is absolutely essential. If the device is to be automatically controlled and powered; a circuit, power source, and controller must be implemented. The overlying specification that must be met in order for the ET device to insure the safety of the neonate is the control of the blood volume flow rate. As discussed in earlier sections of the report, this is a critical specification that must be met. The component of the system that makes this happen is the microprocessor. The other electrical components exist to provide support, or input to the microprocessor (except the buzzer alarm which serves as its own engineering specification). All of components used in the electrical/circuit subsystem have been documented and discussed in the parameter analysis section of the report. Therefore, this subsection of the final design description will take a function-based approach.

User Input Devices: The automated ET device was designed to accommodate the various needs of the neonate and the medical personnel. To oblige to the needs of the medical personnel, the automated ET device must be compatible with an assortment of syringes ranging from 20 to 25 mL in volume. To insure the safety of the neonate, the ET device must be capable of operating at various volume flow rates. The ET device will also be required to stop when a specified volume of blood has been transfused. In order to adapt to all of these variables, the automated ET device must have a user input to inform the ET device of the necessary parameters. From the user input, the ET device will adjust accordingly.

The user interface will consist of three rotary encoders and a LCD. The three rotary encoders will allow the medical personnel to input the necessary system parameters. The three encoders and the LCD will be easily accessed by the medical personnel. The three encoders and the LCD will be soldered onto the top of the circuit board with the LCD screen just above the three rotary encoders as depicted in Figure 52 on page 51.
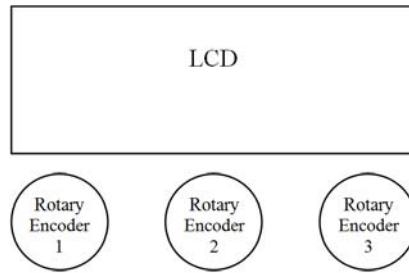
Figure 52. User interface as it appears on the circuit board.

The three rotary encoders will be placed a sufficient distance from each other to insure enough room for the medical personnel to turn the encoders. The encoders can either be spun or pressed by the user. Each encoder can input multiple user inputs; however, the input sequence is determined by the programming of the microprocessor. The rotary encoders will be programmed as follows.

Rotary encoder 1 will accommodate for various size syringes. Syringes with a 20 to 25 mL volume have varying plunger lengths depending on the manufacturer of the syringe, which influences the distance the linear actuator needs to actuate. If the linear actuator moves too far out or too far in, it may cause the syringe to fail; therefore, causing the entire system to fail. In order for the automated ET device to adapt to various syringe sizes, the minimum and maximum position of the linear actuator must be defined. The user will be prompted by the LCD screen to set the minimum position of the actuator. The user will then turn the rotary encoder clockwise (the linear actuator is moving in the outward direction) or counterclockwise (the linear actuator is moving in the inward direction) until the plunger is fully extended from the syringe. The user will then press rotary encoder to set the given position. The user will then be prompted by the LCD screen to set the maximum position of the actuator. The user will then turn the rotary encoder as previously stated until the plunger is fully compressed into the syringe. The user will then press rotary encoder 2 inward to set the given position. The cycle will start by ejecting blood from the neonate; therefore, the linear actuator will be in the correct starting position for when the process begins. Rotary encoder 1 will act as a pause/resume function while the ET is in process; therefore, the neonate can receive the necessary nutrients, such as calcium. Rotary encoder 1 will just need to be pressed once during the ET process to pause. The LCD screen will then display that the process has been paused. Rotary encoder 1 will then need to be pressed again to resume the ET process.

Rotary encoder 2 will maintain two functions: adjusting the volume flow rate and inputting the diameter of the syringe. The current volume flow rate value will be displayed on the LCD screen for the user's reference. In order to adjust the volume flow rate, the user will rotate the encoder clockwise to increase the volume flow rate and rotate the encoder counterclockwise to decrease the volume flow rate. As the user rotates the encoder, the volume flow rate on the LCD will change values for the user's reference. The user will then press rotary encoder 2 for at least 3 seconds and then release the rotary encoder. The user will then be prompted to input the inner diameter of the syringe. The syringe diameter will be displayed on the LCD screen; therefore, the user will rotate the encoder in the clockwise direction to increase the diameter of the syringe or in the counterclockwise direction to decrease the diameter of the syringe. The user will press the encoder inward to set the diameter. The inner diameter of the syringe is an important dimension for the automated ET device to acknowledge, because this will be used in determining the total blood volume that was exchanged.

Rotary encoder 3 will allow the user to alter the total blood volume to be infused into the neonate. The total blood volume to be infused will be displayed on the LCD; therefore, the user will either rotate the encoder clockwise to increase the total blood volume transfused or rotate the encoder counterclockwise to

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decrease the total blood volume transfused. The rotary encoder should be pressed inward to start/stop the process.

Flow Rate Control with a Position Schedule: Control of the blood volume flow rate will be conducted with a position schedule. When the user inputs a desired blood volume flow rate and syringe diameter into the microprocessor, it calculates the necessary linear velocity of the actuator shaft. With the minimum and maximum positions of the actuator set prior to start, the microprocessor determines where and when exactly the shaft should be between the minimum and maximum set points to achieve the desired blood volume flow rate. This is known as a position schedule.

 The linear actuator is equipped with a built-in potentiometer that detects position of the shaft, which is fed back to the microprocessor. If the shaft is detected as being at a position that is ahead of schedule, the microprocessor pauses the motor and the shaft movement until the schedule catches up to the shaft position. If the shaft is detected as being at a position that is behind schedule, instead of speeding the motor up, the microprocessor just shifts the schedule back to coincide with the current schedule.

 It operates this way because being ahead of schedule (having a higher than desired flow rate) is an issue, while being behind schedule (having a lower than desired flow rate) is not so much. This simple schedule method eliminates the need for any continuous feedback control of the motor speed. At any given time during operation, the motor is either operating at a nominal speed, or it is paused.

Process: The device will permit medical personnel to leave the neonate unattended while the device is transfusing blood into and out of the neonate. Since the medical personnel will not be constantly monitoring the neonate, the LCD will output the total amount of blood that has been transfused at that given moment, as well as the volume flow rate. This will provide the medical personnel with an update on the progress of the transfusion. A user manual will be provided to assist the medical personnel with the setup, as well as the operation of the device.

The automated ET device will calculate the total blood volume transfused (V_{tot}) into the neonate by the following equation:

$$V_{tot} = \frac{\pi}{4}(ID)^2 x \text{ (Eq. 13)}$$

where *ID* is the inner diameter of the syringe, which was input into the user interface, and *x* is the total distance the linear actuator traveled in the forward direction, which is determined by the potentiometer. Once the total blood volume transfused meets the value input by the medical personnel, the process will automatically stop. The automated ET device will start to buzz and the LED will shine red to get the attention of the medical personnel.

If a malfunction occurs before the specified blood volume is transfused, the automated ET device will react in the following way. The automated ET device will start to buzz and the LED will shine red to get the attention of the medical personnel; however, the LCD screen will display the word "MALFUNCTION." The primary malfunction that will be detected is leakage. A leak will cause a change in pressure within the system, which will cause a change in the position schedule. This change will be detected by the microprocessor and the microprocessor will process it as a malfunction.

Tubing Subsystem: The tubing subsystem is the connection between the blood reservoirs, the syringes, and the neonate. It is the pathway for the blood. The pathway that the blood takes from fresh blood bag, to syringe, to neonate; and waste blood from neonate, to syringe, to waste bag, is thoroughly described in the preceding section *Process Description*. This section is fairly synonymous with the parameter analysis

section of medical components on p. 42. The reason for this is that the components for this subsystem were chosen mainly on a function basis. For example, 20 and 25 mL syringes are the most readily available syringe at KATH, therefore they shall be the primary syringes that this device is designed for.

While the tubing and its related components such as the one-way valves is considered its own subsystem, its design has already fully been described in the *Process Description* section. Description of the individual components can be found the *Parameter Analysis of Medical Components* Section. Therefore, we refer the reader to those two sections to understand the final design of this subsystem.

Prototype Description

The purpose of the prototype that will be available at the Design Expo on April 16 is to prove the most important elements of the final design. For the most part, the prototype will be highly similar to the final design because of the large quantity of commercially-made components that the final design is expected to contain. The slider, holding block, brace, and housing container will be the only components not acquired through suppliers. Therefore, the material difference between these fabricated parts will be the biggest difference between the prototype and the final design. Because of this fact, the operation of the device at the design expo will be staunchly similar to the how the manufactured device is expected to work in the field.

A description of the differences between the prototype and final design will be discussed. For an evaluation of how the prototype will prove the most important elements of the final design, despite the few differences between the two, please refer to the validation section of the report.

Prototype/Final Design Differences

Materials: The material that the slider is made out of will be different for the prototype and the final design. The prototype slider will be made out of ABS plastic because it will be rapid prototyped with 3D printing technology. The 3D printing supplier that will be used utilizes ABS plastic in their processes. The slider final design will be machined out of PVC material. The use of ABS plastic in place of PVC poses no inconsistencies with the final design material of PVC because the slider does not experience any stresses that come close to the yield strengths of either material. ABS plastic is also just as manufacturable as PVC.

Power Supply: The power source for the working prototype will be 120 V AC, while the final design is compatible with both AC and DC power sources. Also, while the prototype will be tested in the United States where a 120 Hz AC is used, a 220 Hz AC adaptor must be used to make the final design compatible in the intended location of Ghana.

Protective Casing Material (Walls and Lid): Because it was the most convenient material in the machine shop and it was perfect for laser cutting, we used plexi-glass as the material for our overall system casing and lid. However, plexi-glass is not an available material in Ghana, so for our final design, we recommend that this casing and lid should be made out of transparent PVC instead.

Circuit Board Casing: To protect the circuit board and the user interface from debris from the outside or from blood possibly coming into contact with it from the inside, an additional casing should be made for the final design specifically for this electrical component in addition to the overall casing to surround the entire device. The circuit board would be the most difficult component of our device to replace if it was ever damaged, so the final design should definitely include some kind of additional transparent PVC casing to surround its entire body. The material should be transparent PVC because it is conveniently available in Ghana, and because the interface would still be visible from the outside. A slit would need to

be manufactured into this electrical component casing to still allow the user input knobs to be accessible from the outside, however.

Circuit Board Placement: For our prototype, the circuit board was secured into the slot cut into the side wall of the plexi-glass casing near the sliding mechanism. This was probably not the best spot for the circuit board to be placed as it is in very close proximity to many moving parts. For the final design, we would recommend changing the circuit board location to the opposite side wall of the casing, and towards the end of the wall where the linear actuator is. There is more free space near the linear actuator end, so the circuit board would have less of a chance of coming into contact with moving parts if placed there instead of where it was for the prototype.

Track Casing: Along with manufacturing a casing for our electrical component, an additional casing should be made to protect the stainless steel, ball-bearing track that slides the slider back and forth and allows the plunger movement in and out of the syringe. This casing wouldn't necessarily need to be transparent, but there should be a casing made for this for the final design so that no blood could accidentally come into contact with the sliding mechanism. If this was to happen, corrosion could build up in the track and the slider might eventually stop sliding correctly. A recommended material for this casing would be the same PVC used to manufacture the syringe holder, or the same transparent PVC used for the final design outside casing.

Tubing Size: To ensure that we had a working prototype in time for the Design Expo, we used regular tubing purchased from Home Depot to test our system with. This tubing fit over both the one way valves and the y-connectors nicely, so it worked for functionality purposes. However, for the final design, we recommend using 2.5 mm inner diameter medical tubing instead of standard tubing purchased at a hardware store. This 2.5 mm inner diameter medical tubing is designed to be more air-tight than the hardware tubing, and this closer connection would prevent air bubbles from entering the system, thus providing a safer device overall. We should note that smaller one way valves than the ones used for the prototype would need to be used for the final design to be compatible with 2.5 mm inner diameter medical tubing. The one way valves used for the prototype are too big to fit inside the medical tubing. The y-connectors used for the prototype will still fit with the 2.5 mm inner diameter medical tubing, so those would not need to be changed.


Manufacturing, Fabrication and Assembly Plan

The following section will provide the fabrication and assembly process utilized to create the function prototype. The Bill of Materials is included in Appendix B for reference.

Manufacturing Plan

Syringe Holder

*Syringe slots***:** The syringe holder will be made out of PVC from a blank with dimensions of 5.00" x 1.50" x 2.50". Using the band saw, two vertical cuts will be made 1.80" away from each edge to a depth of 0.575". The work piece will then be placed at a 45° to the band saw and 1.225" away from the edge. A cut will be made into the work piece until the saw reaches the intersection of the first vertical cut (a cutting distance of 0.813"). Another cut will then be made, mirroring the first cut about the original vertical cuts. The layout of how the syringe holder should be fabricated is depicted in Figure 53 below.
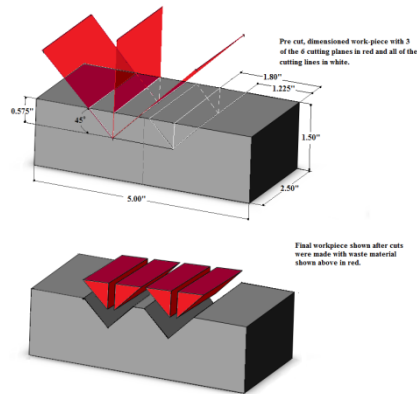
Figure 53. Layout of how the syringe holder should be cut.

These slits need to be centered in order for the syringes to lie straight with the other components, however the angle of the cut can vary up to 2 degrees as long as it's consistent throughout.
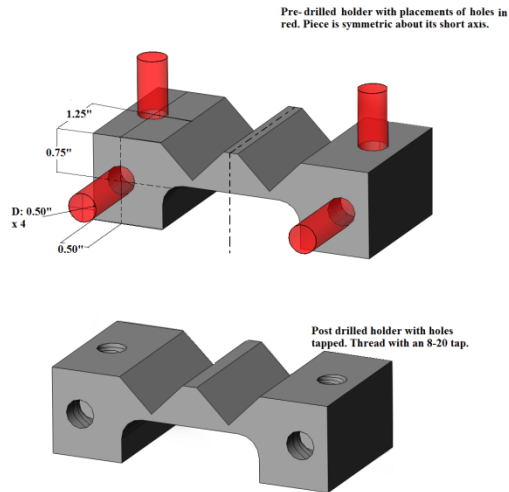
Track hole: After the syringe holder is cut with a band saw, it needs to have a piece from the bottom milled out. Using a ¾ in. diameter end mill bit at a speed of 800 RPM, make an initial cut at 2.50" from the side and the center of your mill bit 0.425" from the bottom as depicted in the Figure 54 below.
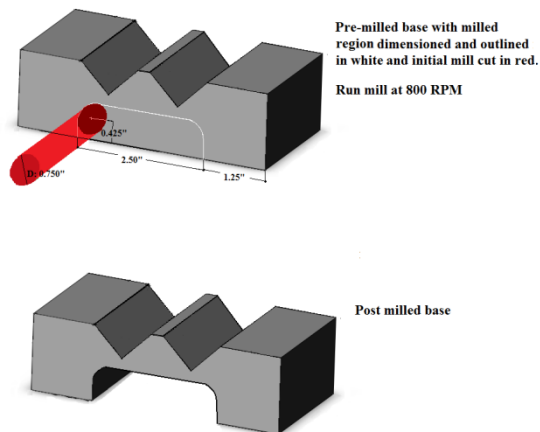
Figure 54. Layout of how syringe holder should be milled.

This size of this space is not that critical. It just needs to allow the track to slide underneath it.

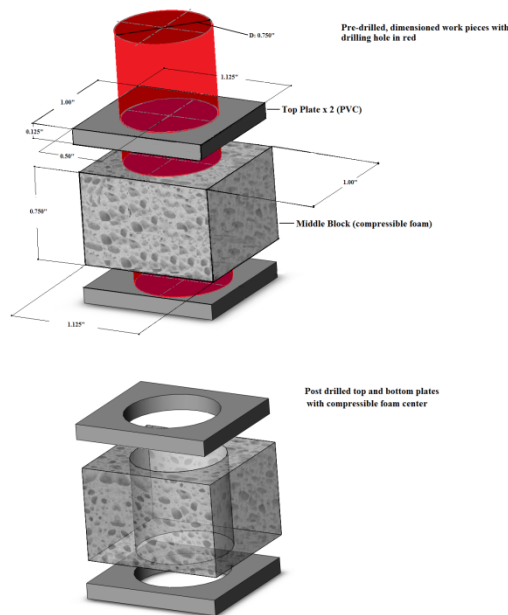*Post placements:* The syringe holder will need four holes in order to place the posts which will hold the top plate and the collar lock. Each hole will be 0.50" in diameter and be threaded with an 8-20 tap. The holes are 0.50" in from one side and are symmetric about the short axis of the piece. The dimensions of where to drill and tap are shown in Figure 55 below.

Pre- drilled holder with placements of holes in red. Piece is symmetric about its short axis.

1.25"

0.75"

D: 0.50"
x 4

0.50"

Post drilled holder with holes tapped. Thread with an 8-20 tap.

**Figure 55. Layout of how syringe holder should be drilled and tapped.**

Top Plate: The top plate will be composed of two separate parts, the feet and top face. Two identical feet will need to be created, made up of three parts; two PVC plates and a middle block made up of compressible foam. The top face is made up of PVC

***Feet*:** To create the feet, take two blocks of PVC measuring 1.00" x 0.125" x 1.125" as well as a block of compressible foam measuring 1.00" x 0.750" x 1.125". Place the compressible foam in between the two pieces of PVC. Drill a hole with the center 0.50" from the long end and centered on the short axis. Drill through all of them with a 0.750" diameter drill bit as seen in Figure 56 below. Do this process twice to make 2 separate feet.

Pre-drilled, dimensioned work pieces with drilling hole in red

D: 0.750"

1.125"

1.00"

0.125"

0.50"

Top Plate x 2 (PVC)

0.750"

1.00"

Middle Block (compressible foam)

1.125"

Post drilled top and bottom plates with compressible foam center

Figure 56. Layout of how the pieces to a foot should be drilled

***Top Face*:** To create the top face, take a piece of plexi glass measuring 3.00" x 1.00" with a thickness of 0.250". The thickness (0.250 inches) can vary +/- 0.05 in. The final plate is shown in Figure 57 below.
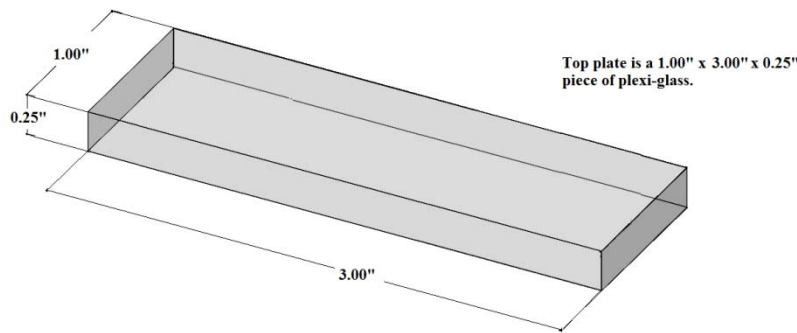
**Figure 57. Layout of how top plate should be drilled.**

Locking Plate: The locking plate will be created from plexi glass cut from a laser cutter. Its overall dimensions are 5.00" x 1.25" with a thickness of about 0.22". A rectangular hole with dimensions 2.50" x 0.50" centered at the bottom is cut from piece. Two holes with diameter 0.75" each centered vertically on the plate with each hole's center placed 0.50" from either end. The final piece would be symmetric about its short axis as depicted in Figure 58 below.
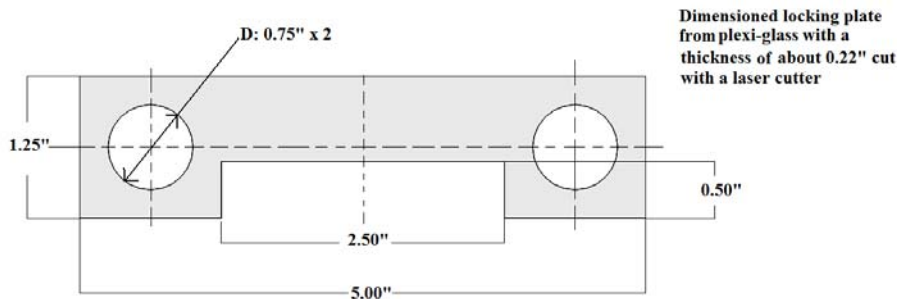
Figure 58. Layout of the locking plate's final dimensions.

Insure that the holes are can fit around the holes drilled into the back of the syringe holder.

Track: The commercial drawer slider will need to be cut so that our design stays consistent with our minimum size constraints. Using a band saw, two cuts that are perpendicular to the track's axis of motion will be made. One of the cuts will need to be made at a distance of 6.00" from the back of the slider. The other cut will need to be made at a distance of 6.50" from the front of the slider. The layout of how the track should be cut is depicted in Figure 59 below.
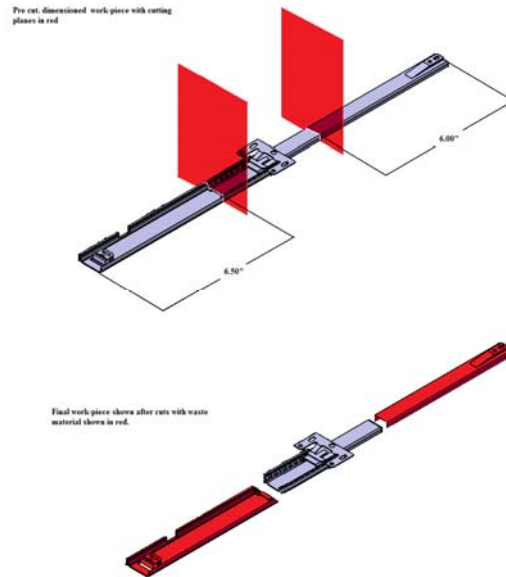
Figure 59. Layout of how the track should be cut.

The track needs to be able to slide a minimum of 4.00". Larger than 4.00" would also work, but 4.00" is the lowest one can go for a fully functional assembly.

Slider: Due to the fine tolerances needed for the slider, creating it in the machine shop would have been extremely difficult and impractical. Therefore, in order to accomplish this fine detail, 3D printing in the Duderstadt Center will be used to create it. Though this method will slightly increase cost, we feel the precision it will give our slider will be worth the extra expense. Also, since the total volume is only $1.33 \cdot 10^{-4}$ m^3, the 3-D printing will actually be quite inexpensive.

Container Walls: The container walls are to be made from plexi-glass with a thickness of about 0.22". Due to the walls being put together with multiple box joints, a laser cutter is the quickest and most accurate means to make the walls of the container. The dimensions of all four final walls are shown below in Figure 60. One would have to upload these designs into a CAD software that works with a laser cutter such as BOBCAD™.
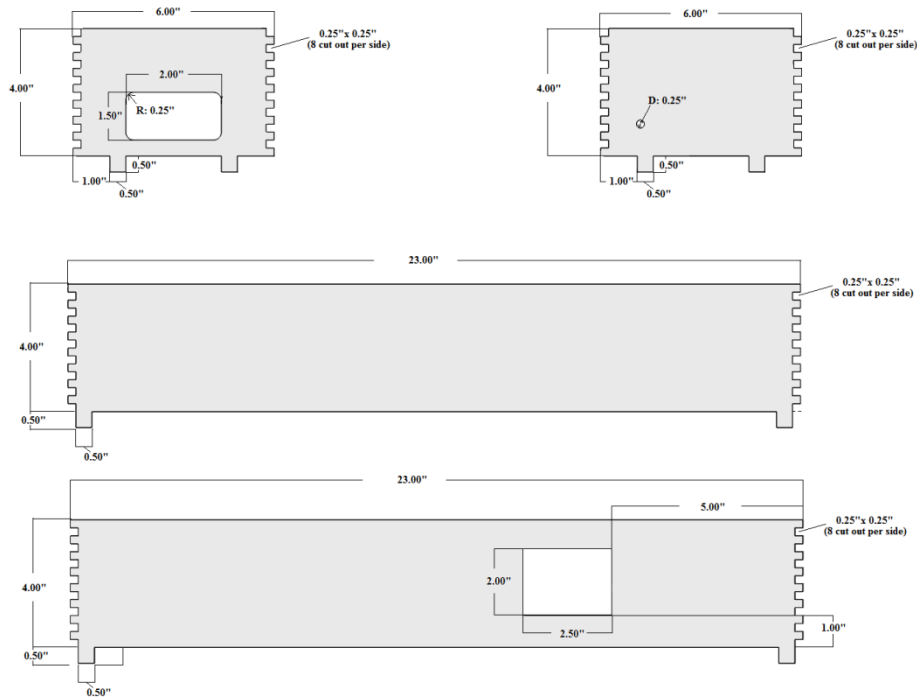
Figure 60. Layout of how walls would look after cut.

Base Board: The base board will be made out of wood and be 6.50" long by 23.50" wide and at least 0.50" thick. Eight separate, rectangular holes (all measuring 0.25" x 0.50") will be cut into the board. They will be needed for the peg placements for the walls and create a concentric rectangle boarder within the board measuring 6.00" x 23.00" as seen in Figure 61 below.
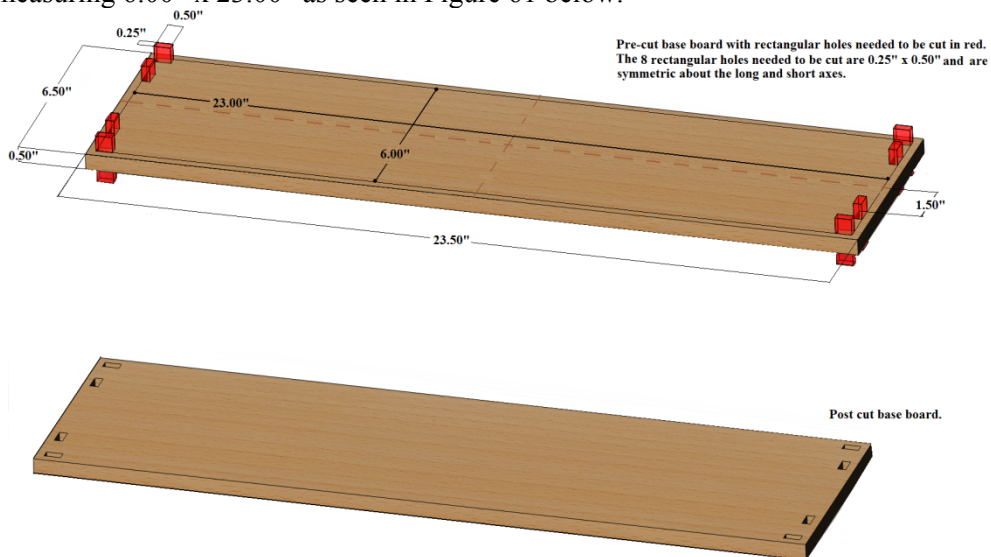

Figure 61. Layout of how base board needs to be cut

The holes need to be totally enclosed within the board. A recommended strategy would be to drill a small hole where the center of each rectangular will need to be and then put together a saber saw within the hole to complete the rectangular cut. These holes need to be fairly precise, due to the walls, which were machined by the laser cutter being put into the slots. A tolerance of + 0.005" could be tolerated.

Fabrication Plan

Top Plate: The top plate consists of two feet (Figure 55, p. 56) and a top face (Figure 56, p. 56). Place the feet under the top face with a 0.125" overlap. Press all parts together securing all parts with appropriate super glue as shown in Figure 62 below. Allow the glue to set before placing onto other parts of the assembly.
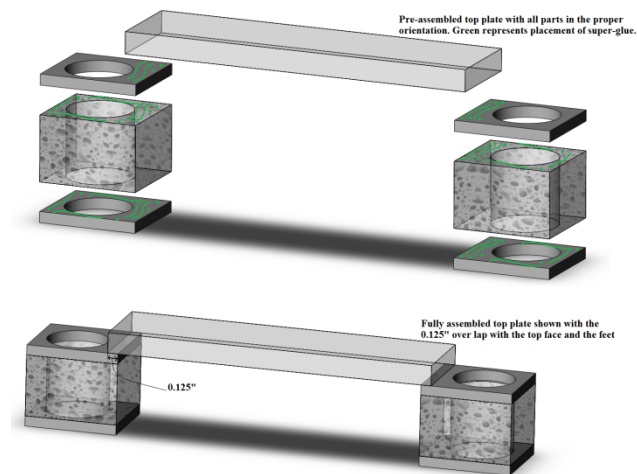
Figure 62. Layout of how feet and top face come together to form the top plate.

Syringe holder and posts: The posts that will hold the top plate and the collar plate will be screwed into the tapped holes. Screw four 2.00" long, threaded posts with a 0.50" diameter into the four holes (posts have an 8-20 pitch). To create the posts, cut the heads off of bolts with the proper diameter and threading. The syringe holder with the posts is shown below in Figure 63.
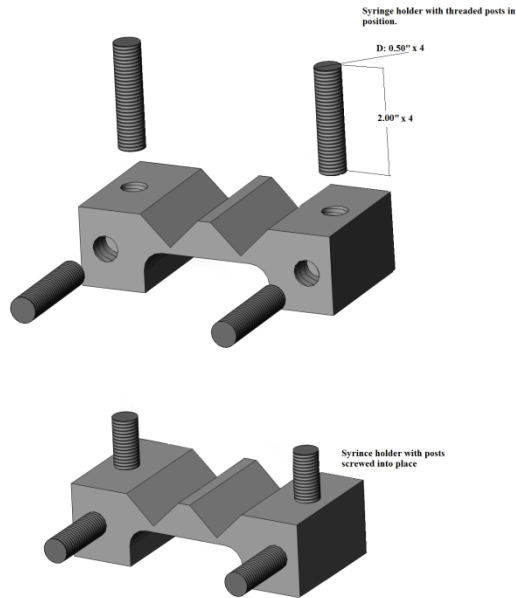
Figure 63. Layout of syringe holder with posts.

Fastening components on base board and track: There are three components that need to be fastened to the base board; the syringe holder, the track, and the bracket for the linear actuator. All three components will be fastened by 0.50" long screw with 0.25" diameters. The screws securing the track and the bracket go down into the board, whereas all others come up in the board. All holes are symmetric about the long axis of the board. The slider is also fastened to the track with these screws. Pre-drilling with a 0.125" diameter bit is recommended. The full layout of where to drill and place all components is shown in Figure 64 below.
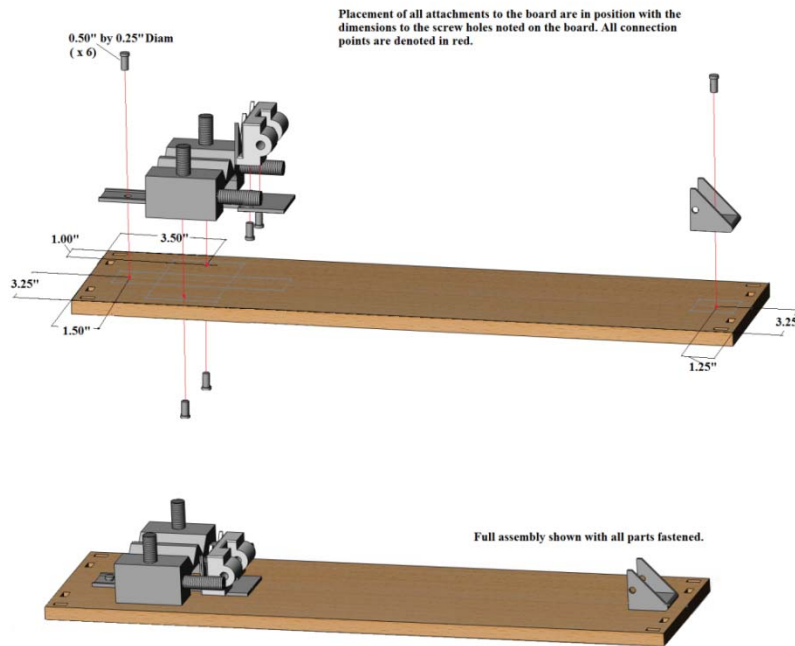
Figure 64. Layout of all components needed to fastened onto the board

Collar Lock and Top Plate: The collar lock and top plate need to be put onto the assembly. The top plate sits on the vertical posts whereas the collar lock slides onto the horizontal posts. It's recommended to keep the nuts on the posts after the plate and lock have been put onto the posts to keep them where they need to be. The full assembly can be seen below in Figure 65.
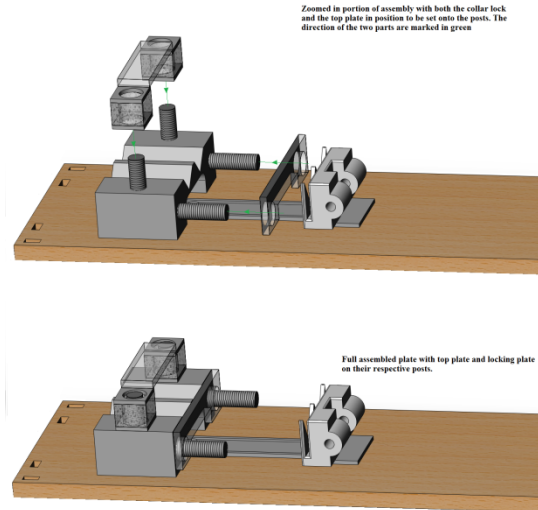
Figure 65. Layout of collar lock and top plate in position.

Walls: The walls are the last of the fabrication. Fit the walls together so they form a box. Adhere the seams with acrylic glue. Once acrylic glue is set, place walls onto baseboard in appropriate slots. Make note of the orientation of the walls. Ensure that the long wall window is closest to the circular hole as shown in Figure 66 below.
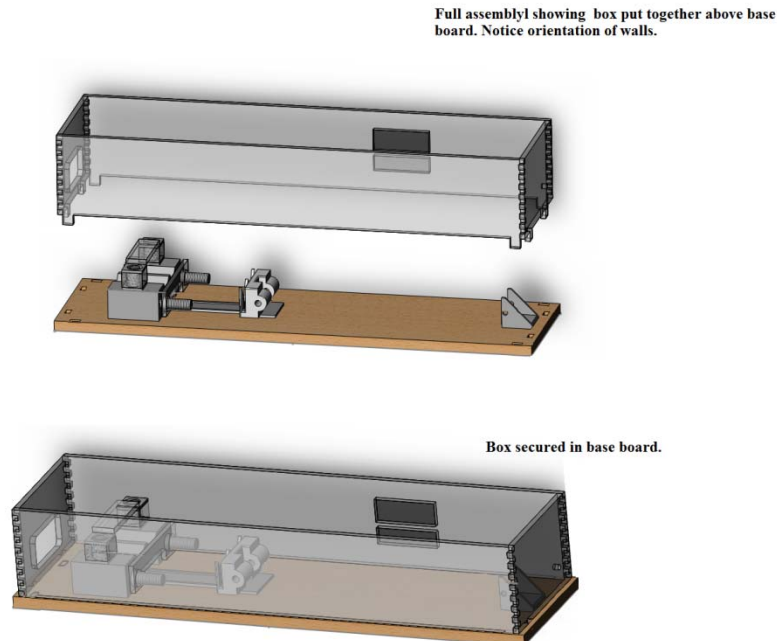
Figure 66. Layout of walls surrounding the full assembly

Validation Results

To validate a working prototype, assembly of the device with all of its components was first required. Once assembly was complete, system testing began and several different experiments were conducted to verify that the device met all of our key customer requirements and engineering specifications. Descriptions of the validation tests and results are described below.

Before we began running our device, we first tested our one way valves to ensure that fluid was restricted to only flow in the one direction that it was supposed to once the valves were in place. For the test, we attached the one way valves to our 2.5 mm inner diameter tubing and manually tried pumping water through the tube both ways using a syringe to make sure that that the water could only flow through the valve in one direction, but could not flow through the valve in the other direction. Once we confirmed that the one way valves functioned properly, we carefully secured the tubing with all attached one way valves to our syringes, our y-junctions, and our IV bags to complete the flow system assembly of the device.

Following the validation of our one way valves and tubing, we tested the time it took to prepare the system. This preparation time consisted of securing the two syringes to the device, inputting sample desired values for the volume flow rate, total volume to be transfused, and inner diameter of the syringes into the user interface, and prepping the tubing with water to act as blood to eliminate any air bubbles from the tubing before attempting to run the system. Syringes of size 20 mL and 25 mL were used at different times during experimentation to ensure that the device was compatible with multiple size syringes and that engineering specification was met. To test the preparation time, we gathered six random participants. Each participant took two turns trying to set up the system. For the first attempt, each of the six participants was given personal help by one of our group members to set up the system. On average, the setup time for this test took approximately 16 minutes and 16 seconds. For the second attempt, each participant had to set up the system using only the instruction manual; no help from any of the group members was given. On average, the setup time for this test took approximately 19 minutes and 17 seconds. A graph of the experimental results can be seen below in Figure 67. Both tests resulted in an average setup time of less than the desired 20 minutes, thus ensuring that the setup time and ease of use engineering specifications were both satisfied.

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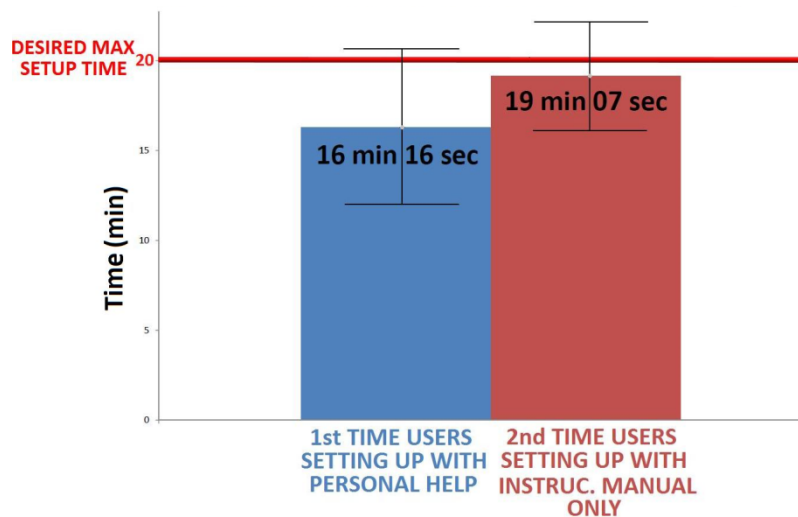
Figure 67. Average setup time of 1st time users with personal help compared to 2nd time users with the instruction manual only.

To have a better understanding of fluid flow throughout the system, we used red and blue colored water for all system tests to act as the blood coming out of the neonate, and the blood going into the neonate, respectively. We also used three IV bags for our testing: one bag was filled completely with red water to act as the fresh blood bag, one bag was left empty to act as the waste blood bag, and the last bag was filled completely with blue water to act as the neonate.

Once the preliminary tests and measures were taken, it was time to power the system to begin testing the functionality of the device as it was running. The linear actuator motor was driven by our AC adaptor to ensure that our device functioned with AC power as required. Because an adjustable and consistent volume flow rate is so important to the proper functionality of our system, we conducted several tests to ensure that the rate stayed consistent with the input value, and that the system could run properly at volume flow rates between 0 and 10 mL/min. For tests running at 2,4,6,8 and 10 mL/min, we manually recorded how long it took the syringe to displace 15 mL of fluid. After calculating the measured volume flow rate, we compared it to the desired volume flow rate. For each of the 5 different flow rates tested, an accuracy of over 99.5% was achieved. A table displaying the full results of the experiment can be seen below as Table 7. This easily satisfied the adjustable volume flow rate engineering specification. We would like to note that it is reasonable to assume that the system had a constant flow rate at all times because the linear actuator's potentiometer is constantly detecting position to ensure a steady volume flow rate.
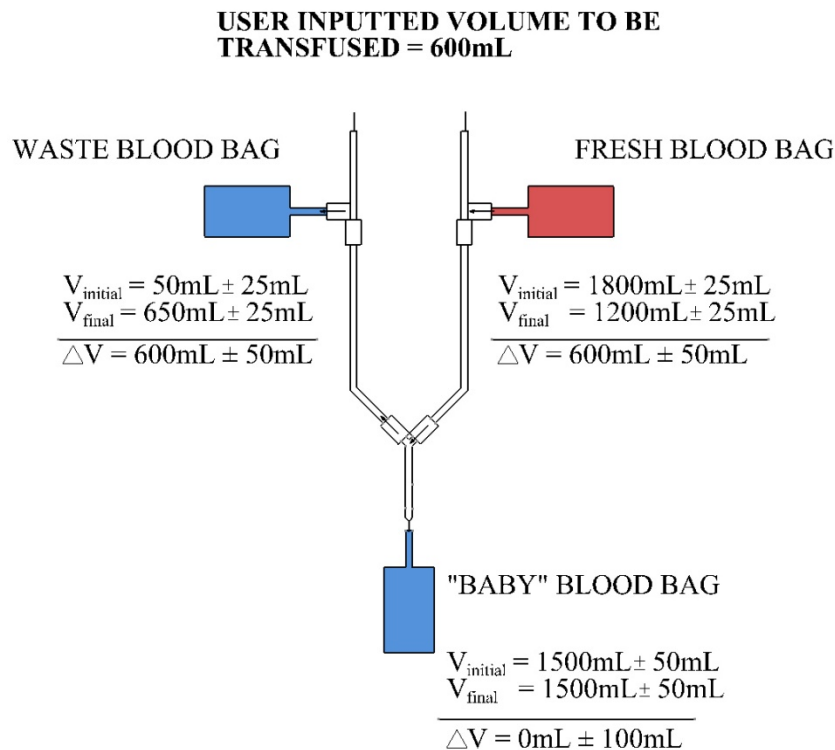
Table 7. Flow rate validation.

Desired Flow Rate (mL/min)	Measured			% Accuracy
	Vol / Stroke (mL)	Time/stroke (min sec)	Flow Rate (mL/min)	
10	15	1 min 31 sec	9.89	98.90%
8	15	1 min 53 sec	7.96	99.56%
6	15	2 min 30 sec	6.00	100.00%
4	15	3 min 46 sec	3.98	99.56%
2	15	7 min 32 sec	1.99	99.56%

To validate that the total volume to be transfused was correct, we ran the system twice for a full two hour cycle at a volume flow rate of 10 mL/min. For these two tests, we first filled the fresh blood IV bag with 1800 ± 25 mL of red water. We then filled the waste blood IV bag with only about 50 ± 25 mL of blue water, and finally the IV bag representing the neonate was filled with 1500 ± 50 mL of blue water. After both two-hour tests, the new total volumes from the bags were recorded. For both tests, the fresh blood bag was reduced by 600 ± 50 mL of water, the waste blood bag was increased by 600 ± 50 mL of water, and the bag representing the neonate remained at a constant volume. A picture clearly depicting these results can be seen in Figure 68. Validating the amount of blood displaced in the fresh blood bag, the waste blood bag, and the "baby" blood bag by color pigment and volume change. From these results, we concluded that the total volume to be transfused was correct for each system test. For further verification that fluid flow was correct going into and out of the bag representing the neonate, we witnessed a color change near the end of the bag where the fluid was flowing in. The color of the water changed from a solid blue to a more purple color as the "fresh blood" red water was injected and mixed with the old, "waste blood" blue water. This color change proved that "fresh blood" was correctly entering the neonate throughout the transfusion.

As stated above, numerous tests were conducted where the system was allowed to run through a full transfusion. For each of these tests, once the device had transfused the desired total volume of blood, the microprocessor sent a signal to the motor to shut down. At this shut down point, an alarm went off as programmed to alert personnel that the transfusion had completed. Once the alarm sounded and the transfusion was complete, we visually confirmed that no more fluid was flowing into the end of the bag representing the neonate. From these results, we concluded that our device had a working automatic shut-off function and that corresponding engineering specification was met.

Engineering specifications that were not directly tested were as follows: inexpensive, functions with DC power, and prevents disease transmission. Because of the power supply and linear actuator, our system is more expensive than originally anticipated. However, we feel that sacrificing cost for safety and efficiency is worth the difference. We did not directly test our system to ensure that it functions with DC power because we could not afford purchasing a battery of any kind for our prototype, but this is definitely something that could be tested in further work beyond this semester. Also, we could not directly test whether or not our device would prevent disease transmission; however, we are confident that because the parts of the device that will come into contact with blood are mostly disposable, this engineering specification will be successfully met.

USER INPUTTED VOLUME TO BE TRANSFUSED = 600mL

WASTE BLOOD BAG

$V_{initial} = 50mL \pm 25mL$
$V_{final} = 650mL \pm 25mL$
_____
$\triangle V = 600mL \pm 50mL$

FRESH BLOOD BAG

$V_{initial} = 1800mL \pm 25mL$
$V_{final} = 1200mL \pm 25mL$
_____
$\triangle V = 600mL \pm 50mL$

"BABY" BLOOD BAG

$V_{initial} = 1500mL \pm 50mL$
$V_{final} = 1500mL \pm 50mL$
_____
$\triangle V = 0mL \pm 100mL$

Figure 68. Validating the amount of blood displaced in the fresh blood bag, the waste blood bag, and the "baby" blood bag by color pigment and volume change.

In summary, our device successfully functions as intended. The one way valves allow the fluid to flow properly into and out of the bags and the neonate, and the motor and pumping mechanism allow the transfusion to take place at various, but constant volume flow rates. The desired total volume to be transfused is what gets transfused, and the device is compatible with multiple size syringes. Also, with proper preparation, air bubbles will be eliminated from the system thus eliminating any kind of danger

regarding air embolism. All of these factors combined allow us to conclude that our automated exchange transfusion device functions safely and successfully.

For future validation of our system functionality beyond practice with IV bags and colored water, we recommend animal testing if all safety requirements have been met and the process is allowable. This would be the next step to take before using the system on neonates. We should note that a great deal of further testing would need to be conducted on our device to ensure maximum performance before animal testing could even be considered.

If animal testing with our device was an option to be considered, a full scale process would need to be conducted first to obtain permission for this. A shortened version of the process for obtaining permission is as follows: First, a premarket approval (PMA) would be required from the FDA. The PMA application would need to contain sufficient information to reasonably assure the FDA of the safety and effectiveness of the device. This required information consists of valid scientific data to demonstrate that the device is safe and effective for its intended use. Once the FDA signed off on the premarket approval, a 510(k) application would have to be sent to the FDA informing them of the exact testing procedure to be done on possible animal subjects. If this application was approved by the FDA, subsequent animal testing could then be legally performed.

Discussion

As we stated in the validation section of our report, we are very pleased with the functionality of our device overall. The device successfully meets all of our key engineering specifications. The true strengths of our device lie in its functionality. First of all, the user interface is easy to understand with its directions and the input values remain consistent with the actual values while the system is running. The linear actuator is a perfect motor for creating the linear motion necessary to move the syringe plungers in and out. The syringe holder and top plate designs are more than sufficient for allowing multiple size syringes to be used. The slider mechanism allows for the proper syringe plunger movement. And finally, the one way valves and tubing allow for the correct fluid flow for an exchange transfusion to take place. Given the 400 USD budget we had for the semester, we could not have hoped for a much better end result.

Although we are satisfied with the functionality of our device, there are some things we may have done differently if we could do the project over again along with some weaknesses that could be addressed in the future. First of all, we would possibly try and find a smaller, more inexpensive linear actuator to better fit the size constraints at the KATH hospital in Ghana. The linear actuator we purchased worked perfectly for our prototype needs and it was one of the cheapest we could find at the time, but further research could possibly find an even better product. The linear actuator we purchased was capable of pushing forces well beyond anything we needed it to do, so a smaller scale actuator would definitely be more desirable. Also, the linear actuator was approximately half of the total cost of our device so if that cost could be decreased even a little bit, it would be very beneficial.

Another change we would make to our prototype if we could do things differently would be to add another casing to put around the electrical component. We have an outside casing to surround our entire unit, but a separate casing around the circuit board would be beneficial to eliminate the possibility of fluid ever accidentally coming into contact with the circuitry. Also regarding the circuit board/user interface, we would probably attach that to the top of the casing instead of the side of the casing next time so that it would be more convenient to the user. At this point, the user has to bend down quite a bit to input the values into the user interface and to read the values off the interface while the system is running. As for the outside casing, instead of using plexi-glass, we would probably use transparent PVC if we could make that all over again for a little more added stability.

One more improvement to be made to the final design would be to use actual medical tubing that fits correctly with the y-connectors and one way valves we purchased. Because of time limitations, we used tubing purchased from home depot for testing purposes. This tubing allowed for proper functionality of the system, but medical tubing that would correctly fit with the y-connectors and one way valves would help eliminate air bubbles much better, creating a safer device in case a real blood transfusion was to ever take place.

Recommendations

Although we successfully created a working automated exchange transfusion device, there are still quite a few recommendations we have for Dr. Adabie that could significantly increase the likelihood of this device becoming a real medical device some day. Obviously, we were only given a semester's time to work on this project, and we were working under a limited budget of 400 USD. With a larger budget and more time to work, we feel the in-phase slider could definitely be turned into a realistic medical device. Several recommendations are given in the following paragraphs to make that happen.
The first recommendation we would give would be to find a smaller scale linear actuator that could be purchased in Ghana. The linear actuator is a perfect motor to drive the rest of the device properly, but it is by far the most expensive part of the device. If a smaller, but still effective, similar motor could be used for this device in Ghana, the device would be much more likely to succeed long term. Next, we would recommend a better way to secure the circuit board and user interface. Right now, it is snugly fit into the wall of the casing, but an electrical component with that kind of importance should probably be more securely attached. We recommend using very small screws such as 1 mm diameter screws to fasten the circuit board to the casing wall, and also developing a casing to go around the board to protect it even more.

While manufacturing the prototype, we had to make several compromises because of time constraints and material constraints. Many of the parts were bolted down quickly, and in one instance we had to use super glue to lock the ball-bearing track into place on the mounting board to prevent wiggle. With more time available, we recommend more precise measurements and more quality methods of securing the components to each other and to the mounting board. This would prevent things like slanted bolt attachments, which could result in slipping or wiggling of components later on. More specifically, wing nuts should be used for all attachments that will need to be detached constantly such as the syringe cover plate and the syringe plunger bracket to allow for an easier assembly and setup for the user.

If more circuit boards were to be assembled in order to mass produce this device some day, we would recommend having 4 knobs on the user interface instead of 3. This would make the user input a little easier for the intended user, because one knob wouldn't have to be pushed in to allow for two different functions using that same knob. If this were to happen, the programming would also need to change so that the new fourth knob carried out the same task that the one knob did once it was pushed in to perform its second function.

Conclusions

After the completion of our alpha design, we incorporated several new developments into the design that significantly changed the device. The first new development into the design was the change of the driving mechanism from a gear/crank and connecting rod to a commercial, linear actuator. Using this linear actuator also changed our feedback system. The linear actuator has a built-in potentiometer which will detect position, and the microprocessor will use this position sensing to calculate volume flow rate throughout the transfusion. This differs from our original plan to use pressure transducers to directly detect volume flow rate inside the tubing. Another new development since the alpha design was the

evolution of our sliding mechanism. The new sliding mechanism now has a ball-bearing stainless steel track to glide on that will significantly reduce friction, and a new slider and slider holder design.

These new developments in our design changed quite a few of our engineering specifications. First, we increased the importance of having a design that functions with AC power and decreased the importance of compatibility with DC power. We feel that AC power is more reliable and safe, and should be the primary source for powering our prototype. Because of the evolution in our sliding mechanism, we changed the specification of our device being compatible with various size syringes from 10-25 mL syringes to 20-25 mL syringes. Finally, because of the cost of the linear actuator, we changed the specification of having a device cost less than 100 USD to 200 USD. We felt that because our device could potentially be saving a life, increasing cost was worth the increase in safety and stability. The mobility engineering specification also changed because of the weight of the linear actuator from needing to weigh less than 45 N to 65 N. This is still more than acceptable for mobility purposes.

In order to incorporate the new developments into the slider mechanism and the driving mechanism, and to design for overall compatibility with all of the components of our system, we had to perform parameter analysis for each part of the design. Force analysis was conducted on the slider parts to ensure their stability to prevent buckling or failure once manufactured. Functional analysis was performed on the electrical components of the system before the development of the circuit board and purchasing of the power supply. Experimental analysis was performed on our medical supplies, particularly the syringes, to determine the parameter requirements for our linear actuator. Once this analysis was completed, we moved forward toward the purchasing and manufacturing of all of our system components.

As we have previously stated, we have named our device to be the in-phase slider mechanism. This is an exchange transfusion system utilizing an AC power supply, a linear actuator with a built-in potentiometer as a driving mechanism, a dual-syringe pumping mechanism with a stainless steel, ball bearing track and PVC slider mechanism, a one-way valve configuration, and one IV insertion point into the neonatal umbilical vein. There are some differences between our prototype and our final design, however. As previously stated, our prototype will function strictly with AC power because DC power is not really necessary. However, for our final design, we recommend a power supply that is compatible with both AC power and DC power. The AC power supply for our final design needs to be 220 Hz compatible for the KATH hospital in Ghana, as opposed to 120 Hz compatible for in the U.S. Also, the syringe holder and slider should possibly be made out of other materials aside from PVC and ABS plastic if a different material is as safe, but more convenient to be used for manufacturing these parts in Ghana.

Once we manufactured the mounting board, locking plates, and Plexiglas casing, we attached everything securely onto the mounting board. While the manufacturing was going on, we were also assembling the circuit board and programming it to function as intended to allow for user input of desired volume flow rates, total volumes to be transfused and inner diameters of syringes. Once these tasks were completed, and the rest of the system components were purchased and acquired, we began testing and validating our design. We used red and blue colored water to represent fresh and waste blood, respectively, throughout the transfusion. We also created an instruction manual for setting up the system and performed an experiment using random test subjects to see how long it took them to set up our system both with our help and with just the aid of the instruction manual. The average set up time was less than 20 minutes, so that was desirable according to our engineering specifications.

We also validated the volume flow rate throughout the system for several different flow rates by recording the time passed by for the device to inject or withdraw 15 mL of fluid. All of these tests resulted in a volume flow rate accuracy of above 99.5% so we concluded that the volume flow rate at any given time throughout the transfusion was extremely accurate. To ensure that the total volume being transfused was accurate and that the neonate would be losing the same amount of waste blood as he/she

would be gaining in fresh blood, we measured volume displacement after several full, two-hour transfusions at 10 mL/min each. The accuracy for these was again extremely high.

From these validation tests, we concluded that our device met all of the key customer requirements and engineering specifications, and it functioned properly. The motor provided the correct force, the slider and pumping mechanism pushed and pulled the syringe plungers properly, and the tubing and one way valves allowed the fluid to flow properly. If any of these mechanisms malfunctioned, the validation tests would have failed, and that was never the case. Therefore, we are pleased to say that we successfully developed an automated exchange transfusion device as intended.

Acknowledgements

References

1. WHO. *Country Health System Fact Sheet 2006 - Ghana*. [PDF] 2006 [cited 2009 14 Feb]; Available from: http://www.who.int/whosis/en/.
2. Hillyer, C.D., Luban, N.L.C., & Strauss, R.G. , ed. *Handbook of Pediatric Transfusion Medicine*. 2004, Elsevier Academic Press: San Diego. 394.
3. WHO, *The Clinical Use of Blood in Medicine, Obstetrics, Paediatrics, Surgery & Anaesthesia, Trauma & Burns*. 2001, Malta: World Health Organization. 337.
4. Argarwal, R., Ashok, D.K., Coshic, P., Jain, R., & Jose, B. (2008) *Blood and blood component therapy in neonates*. Indian Journal of Pediatrics **75**, 489-495.
5. Kleinman, S., Petz, L.D., Spence, R.K., Strauss, R.G., & Swisher, S.N., *Clinical practice of transfusion medicine*. 3rd ed. 1996, New York: Churchill Livingstome Inc. 1115.
6. Fasano, R., & Luban, N.L.C. (2008) *Blood component therapy*. Pediatric Clinics of North America **55**, DOI: 10.1016/j.pcl.2008.01.006.
7. Churchill, W.H., Keszler, M., Richardson, D.K., Ringer, S.A., & Sacher, R.A. (1998) *Variations in transfusion practice in neonatal intensive care*. Pediatrics **101**, 194-200.
8. Grindeanu, L.A., Mdad, A., Stevenson, D.K., Widness, J.A., Wong, D.K., & Zimmerman, M.B. (2005) *Reduction in red blood cell transfusions among preterm infants: Results of a randomized trial with an in-line blood gas and chemistry monitor*. Pediatrics, 1299-1306 DOI: 10.1542/peds.2004-1680.
9. Hemlata, V., A. (2008) *Blood component therapy*. Indian Journal of Pediatrics **75**, 717-722.
10. Smith, C.H. (1956) *Transfusions in pediatric practice: Indications and limitations*. Pediatrics, 596-604.
11. DRE. *DRE SP1500 Plus Standard Syringe Pump*. [cited 2009 February 18, 2009]; Available from: http://www.dremed.com/catalog/product_info.php/cPath/45_109/products_id/1659.
12. Tu, H.C., *Exchange Transfusion Machine*, U.S. Patent, Editor. 1984: U.S.

Biographies

The following section provides a brief biography of the mechanical engineering students working on the automated exchange transfusion device for Ghana: Daniel Ciko, Brad Herron, Ashley May, and Devon Stanforth.

Daniel Ciko

Dan Ciko is from Livonia, Michigan, United States of America, North America, Western Hemisphere, Earth, Sun's Solar System, Milky Way Galaxy, Local Group, Virgo Supercluster. He has interest in mechanical engineering because he wants to make the Earth a better place for humans to live in by manipulating the god-given raw resources left here for us. A humanist at heart; he seeks to combine the cold, ruthless calculations of an engineer with a curious, empathetic worldview, to solve problems that are both technological and human in nature.

Dan interned at the Toyota Technical Center in Ann Arbor last fall, and would like to get an internship in either the aerospace or medical technology industry for this upcoming summer. He would like to experience as many industries involving mechanical engineering as he can, before he makes a decision on which one to settle in after graduation. Dan has his sights set on getting a master's degree in mechanical engineering from the University of Michigan and will be applying after this current semester.

Brad Herron

My name is Brad Herron. I am from Clarkston, Michigan. I have lived in Clarkston my whole life. I chose Mechanical Engineering to be my major because I have always loved cars and how engines work. I have not yet had a summer internship since I have attended the University, but I plan to attend the career fair this term and pursue an internship this summer and hopefully extend that work to a full-time position after I graduate next fall. Thermodynamics is probably my favorite topic related to Mechanical Engineering, so I hope to find work where I can use the knowledge I've learned to excel in that field.

I have a girlfriend named Katie who is a sophomore at Albion University. We have been together for about 6 years now, so she is a big part of my life. I have always had a very close relationship with my family as well, which consists of my two brothers, my mom, and I. My dad passed away in September of '07 from Pancreatic Cancer. My older brother lives in California and is a screen writer, and my little brother is a freshman at Western Michigan University. My mom has always been a stay-at-home mom, and I wouldn't have wanted it any other way.

Another big part of my life revolves around sports. Despite being a mechanical engineer and always having a strong math background; truthfully, I would definitely rather work for ESPN than anywhere else. My TV never leaves ESPN, and I am an avid sports fan. I'm not so much into hockey, but baseball, basketball and football I could watch all day. I don't think I could be as happy doing anything else as I would be working as an anchor for ESPN. I majored in mechanical engineering because of the prestige and the money security, but working in sports has always been my dream. In my spare time, I like playing

video games such as Call of Duty and Halo, watching movies from A to Z, and playing with my dog Rocky.

Ashley May

I was born on October 26th, 1987 to my two loving parents. I was second of two children to be born to Kenneth and Sandra May. In 1981, my parents were blessed with my brother John. The four of us lived in a ranch styled home built by my father in Livonia, Michigan, a suburban city (population around 100,000) just outside of metro Detroit. We lifted our roots after thirteen years, and moved to a larger home in the northwest area of Livonia, Michigan. This is where I currently live when I am not attending school.

In the fall of 2005, I commenced my education in the School of Engineering at the University of Michigan Ann Arbor. I am expected to earn my bachelors degree in mechanical engineering in May 2009. I chose to major in mechanical engineering because I have always been intrigued in improving existing mechanisms to the best of their ability, as well as knowing how everything around me functions. During the summer of 2006, after my first year of college, I interned at Orchard, Hiltz & McCliment, Inc., a civil engineering consulting firm located in Livonia, Michigan. In the summer of 2008, I relocated to Greenville, South Carolina to work for General Electric – Energy. I am currently seeking out a fulltime mechanical engineering position. Once I find my calling in the engineering field, I hope to return back to school to get my master's degree.

Devon Stanforth
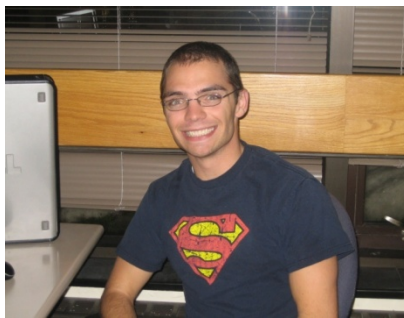
My name is Devon Stanforth and I am finishing up my last semester here at the university. I am a part of the Army Reserved Officer Training Corps (AROTC) here on campus which means right after I graduate in May, I will also commission into the United States Army as a 2nd Lt. working in the Armor Corps. The next four years of my life, I will be commanding platoons of tanks, leading the best group of men the USA has, as well as thoroughly learning these complex mechanical systems. After the Army, I hope to work for General Dynamics (the company that engineers the U.S. military's heavy machinery). As of right now, I am engaged to my High School sweet heart and we plan on marrying next spring.
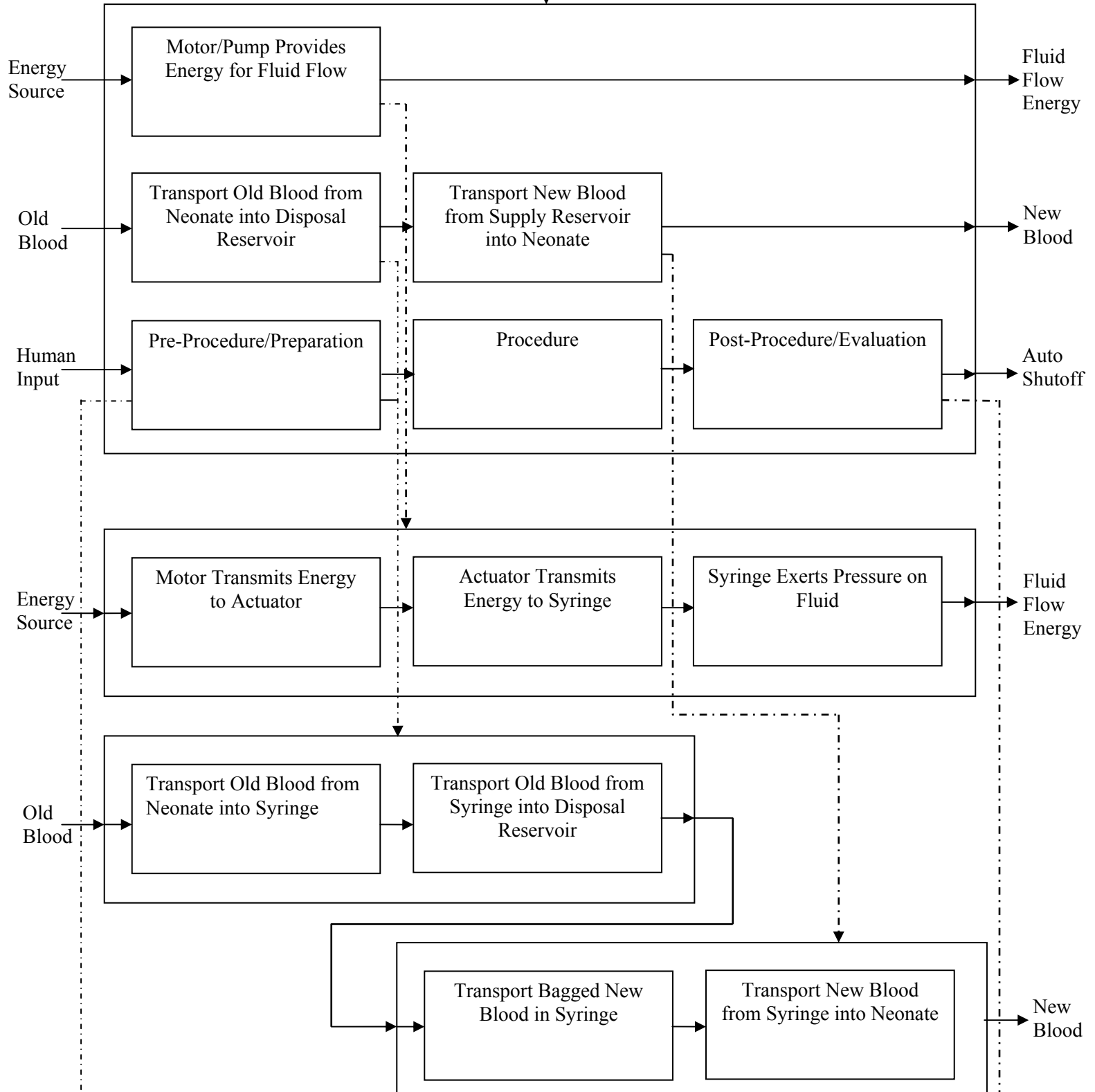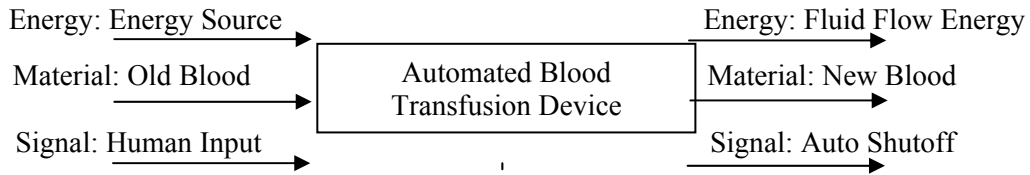
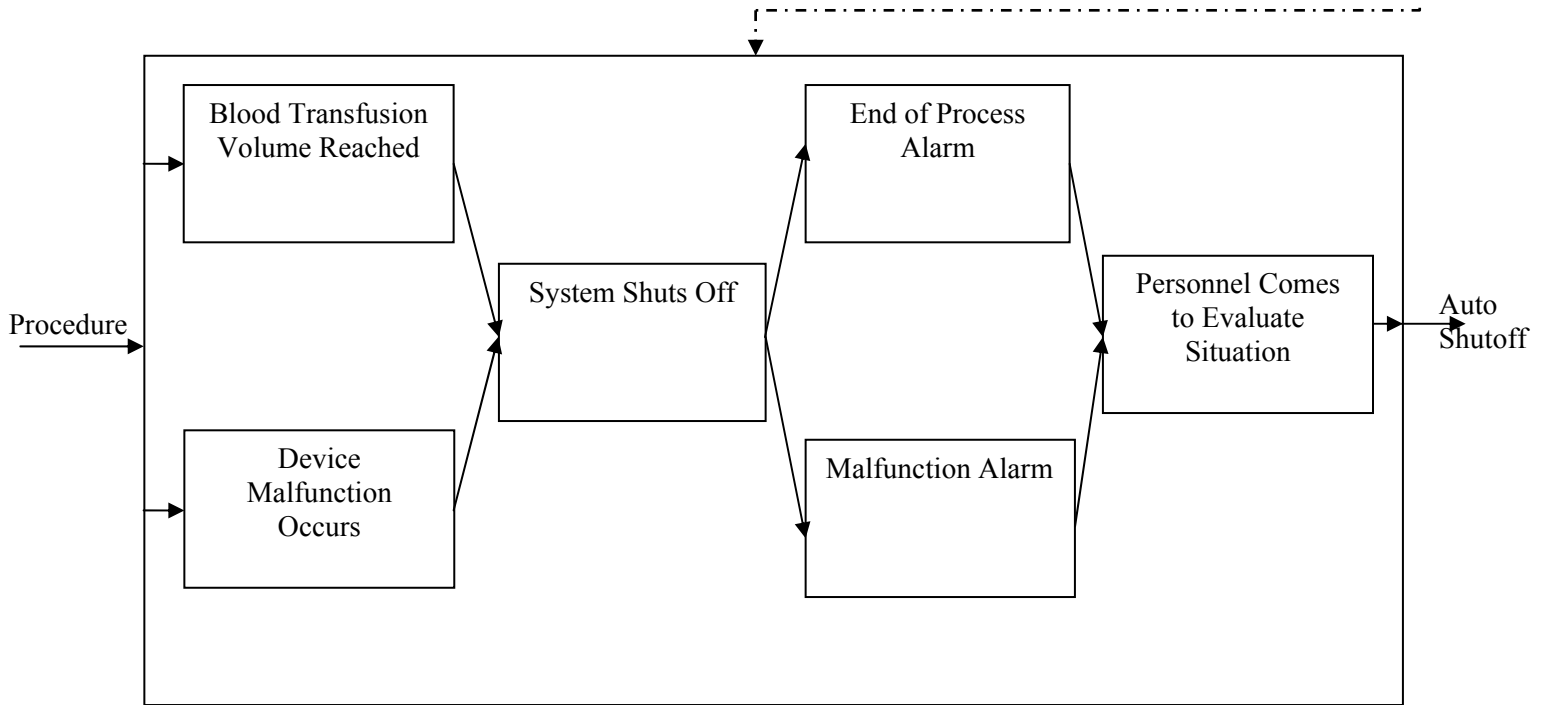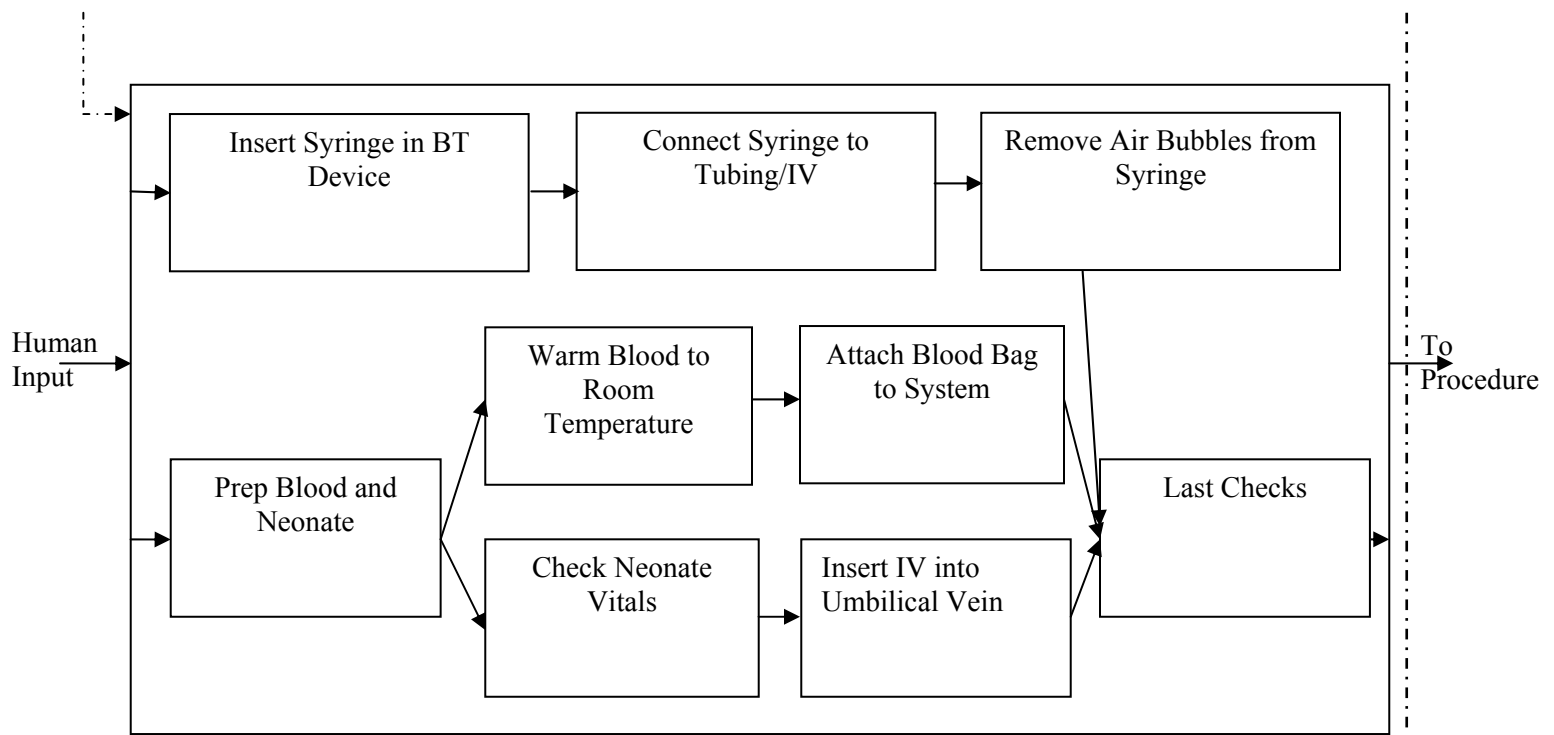**Appendix A-1: Complications of exchange transfusions in neonates [2].**

Infections	Bacterial
	Viral
	Fungal
Metabolic Complications	Hyperkalemia
	Hypocalcemia
	Hypoglycemia
	Hyperglycemia
	Hypernatremia
	Late onset alkalosis
Hematologic Complications	Hemolysis
	Anemia/polycythemia
	Thrombocytopenia
	Neutropenia
	Coagulopathy
	Graft-versus-host reaction
Cardiovascular Complications	Arrhythmia or arrest
	Volume overload
Catheter Complications	Umbilical vein/artery perforation
	Air embolism
	Thromboembolism
	Portal vein thrombosis
	Necrotizing enterocolitis
	Bowel perforation
	Cardiac arrhythmia
Other	Change in intracranial pressure
	Hypothermia/hyperthermia
	Emesis with aspiration

Appendix A-2: QFD Diagram

Technical Requirements (columns):
1. < 0.1 mL air bubbles in syringe
2. Synchronous intrusion/extrusion strokes
3. Device compatibility with syringes between 5 mL and 25 mL
4. Adjustable volume flow rate (0 - 25 ml/min)
5. Meets the Health and Safety Codes of Ghana
6. Audio alarm of 80 dB
7. Audio alarm of 60 dB
8. Spare parts available within 1 week's time
9. 1 IV attachment
10. Setup <5 min
11. Stepper motor
12. Minimum instruction necessary (1 page of instruction)
13. Baby and device must fit in 19x40x22 in. incubator
14. Motor does not exceed 45 dB
15. Device Weight <10 lbs.
16. 220 V 50 Hz compatibility
17. Takes less than 5 min to fully charge mechanism
18. Total cost is less than US $100
19. < 60 seconds to transition from automatic to manual
20. Feedback system
21. 50% disposable parts
22. Microprocessor

Roof (correlation) entries
- Device compatibility with syringes between 5 mL and 25 mL: 3
- Adjustable volume flow rate (0 - 25 ml/min): 1, 3
- Audio alarm of 60 dB: 9
- Spare parts available within 1 week's time: 9
- Stepper motor: -9, -1
- Minimum instruction necessary (1 page of instruction): 9
- Motor does not exceed 45 dB: -3
- Total cost is less than US $100: -3
- Feedback system: 3, 9, -9, 3, -3
- 50% disposable parts: 3
- Microprocessor: -9, 3, -3, -3, 9

Main Matrix

Customer Needs	Cust. Wt	Kano	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22
Adjustable volume flow rate	3				3	9							9											
Alarm device to detect malfunction	1							9	1															1
Alarm device to detect the end of the process	1							1	9															1
Applicable with various syringe sizes	9				9																		3	3
Automatic shutoff at the end of process	9																						9	9
Available spare parts	3									9														
Blood volume control	3																							9
Constant monitoring from doctor is not required	9			9				3	3		1		3											
Ease of assembly/disassembly	9											9	1	9				3						
Ease of sanitizing/cleaning	1						3																9	
Ease of use	9					1							1	9				1					1	
Easily maintained and repaired	3									9		3											3	
Easy transition to manual process	1																				9			
Functions with AC power	3												9						9					9
Functions with battery power	9									1			9							3				9
Functions without the use of electricity	1																				9			
Inexpensive	9									3			9							9				
Instruction manual	9													9										
Minimize sound	1							1	1				1			9								
Minimize the size	1												1		9		9							
Mobile	3												1		9		9							
Operable for both blood transfusion and exchange transfusion	3										3													
Prevents air bubbles	9		9																					
Prevents disease transmission	9																						9	
Safety of neonate	3						1				9							1					9	
Transfuse various types of blood (PRBC, FFB, platelets, etc.)	3						3						3											

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22
Raw score	81	81	90	45	6	38	38	90	36	99	275	243	36	12	72	27	9	108	9	108		245
Scaled	0.295	0.295	0.327	0.164	0.022	0.138	0.138	0.327	0.131	0.36	1	0.884	0.131	0.044	0.262	0.098	0.033	0.393	0.033	0.393		0.891
Relative Weight	5%	5%	5%	3%	0%	2%	2%	5%	2%	6%	16%	14%	2%	1%	4%	2%	1%	6%	1%	6%		14%
Rank	9	9	7	12	21	13	13	7	15	6	1	3	15	18	11	17	19	4	19	4	19	2

Appendix A-3: Functional Decomposition Block Diagram

Energy: Energy Source → | Automated Blood Transfusion Device | → Energy: Fluid Flow Energy

Material: Old Blood → | Automated Blood Transfusion Device | → Material: New Blood

Signal: Human Input → | Automated Blood Transfusion Device | → Signal: Auto Shutoff

---

Energy Source → **Motor/Pump Provides Energy for Fluid Flow** → Fluid Flow Energy

Old Blood → **Transport Old Blood from Neonate into Disposal Reservoir** → **Transport New Blood from Supply Reservoir into Neonate** → New Blood

Human Input → **Pre-Procedure/Preparation** → **Procedure** → **Post-Procedure/Evaluation** → Auto Shutoff

---

Energy Source → **Motor Transmits Energy to Actuator** → **Actuator Transmits Energy to Syringe** → **Syringe Exerts Pressure on Fluid** → Fluid Flow Energy

---

Old Blood → **Transport Old Blood from Neonate into Syringe** → **Transport Old Blood from Syringe into Disposal Reservoir**

Transport Bagged New Blood in Syringe → **Transport New Blood from Syringe into Neonate** → New Blood

Insert Syringe in BT Device → Connect Syringe to Tubing/IV → Remove Air Bubbles from Syringe

Human Input →

Prep Blood and Neonate → Warm Blood to Room Temperature → Attach Blood Bag to System

Prep Blood and Neonate → Check Neonate Vitals → Insert IV into Umbilical Vein

Last Checks → To Procedure

Blood Transfusion Volume Reached → System Shuts Off → End of Process Alarm

Procedure →

Device Malfunction Occurs → System Shuts Off → Malfunction Alarm

End of Process Alarm / Malfunction Alarm → Personnel Comes to Evaluate Situation → Auto Shutoff

Appendix A-4: Brainstorm Concept Drawings

Drawing	Name	Description	Pros	Cons
	Out-of-Phase Slider	Similar to alpha-design. However syringes operate out of phase.	Process time reduced by half. Controllable blood volume flow rate.	Two IV insertion points.
	Gravity-Piston	Weight of waste-blood forces piston down accelerates blood volume flow rate.	No motor needed. Powered by force of gravity.	No control of blood volume flow rate. Must transfer blood from bag to tank.
	Compressor System	Pressure difference mainly drives blood flow rate. Compressor implemented to maintain pressure difference.	Good for adults who can withstand higher blood volume flow rates.	Complex system that still relies on natural flow of blood. No advantage to making the system so complex.
Automated Four-Way Stopcock		Pressure difference mainly drives blood flow rate. Compressor implemented to maintain pressure difference.	Automated. Parts are available locally.	Two IV insertion points.
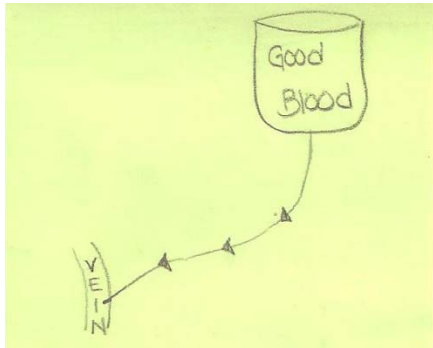
Drawing	Name	Description	Pros	Cons
	Drip Bag	Bag connected to vein. Gravity slowly works on blood and draws it to vein.	Simple. Parts available at KATH. No motor needed.	No blood extraction. No blood volume flow rate control.
	2 continuous electrical pumps	Both the intrusion and extrusion lines have their own electrical pump that pumps blood either into or out of the neonate	Easy to make. Flow rates variable	Purchasing pumps. Contaminating pumps. Two insertion points
	One syringe slider	One syringe is used as the pumping method which is powered by a motor, which puts the fresh blood into the neonate and bad blood into the disposal bag	Automatic. Simple set-up. Available parts.	No way to keep the fresh blood and the bad blood from mixing
	One tank gravity piston	Waste blood weigh forces the piston down, pumping the fresh blood into the neonate	Powered by gravity, no motor needed.	Changing containers for blood. No control. 2 insertion points
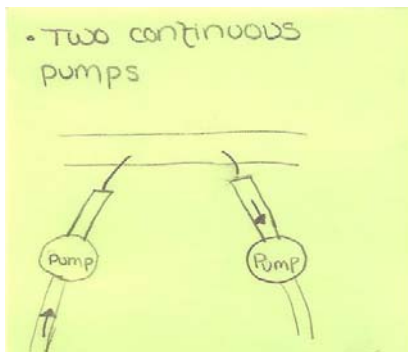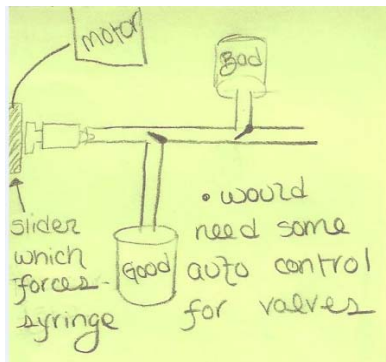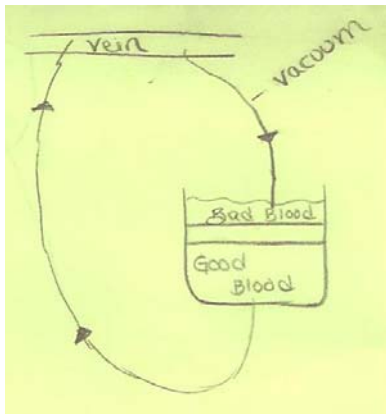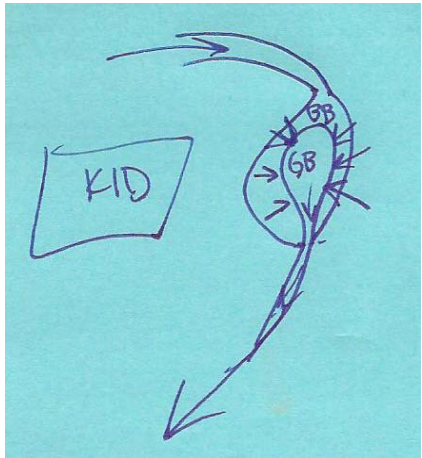
Drawing	Name	Description	Pros	Cons
	Bag Within a Bag	Initially all fresh blood is in inner bag flowing naturally into neonate. Outflow neonate blood surrounds inner bag, compressing it aiding the fresh-blood flow.	Self-powered.	Flow rate is controllable. Bags are custom made. Blood must be transferred from original bag to inner bag. Two insertion points.

Appendix A-5: Pugh Chart

Criteria	Concept 1 *Out of Phase Slider*	Concept 2 *In Phase Slider*	Concept 3 *Gravity Piston*	Concept 4 *Compressor System*	Concept 5 *Automated Four-Way Stop Cock*
Adjustable volume flow rate	1	1	-1	-1	1
Alarm device to detect malfunction easily implemented	1	1	1	1	1
Alarm device to detect the end of process easily implemented	1	1	1	1	1
Available spare parts	0	0	-1	-1	-1
Comfortability of neonate (Number of insertion points)	-1	0	-1	-1	0
Constant monitoring from doctor is not required	1	1	1	1	1
Ease of assembly/disassembly	0	0	-1	1	-1
Ease of sanitizing/cleaning	-1	-1	-1	0	-1
Ease of use (Number of steps after setup)	1	1	1	1	1
Easily maintained and repaired	-1	-1	-1	-1	-1
Functions without the use of electricity	-1	-1	1	-1	-1
Inexpensive	-1	-1	-1	-1	-1
Minimize Sound	0	0	1	1	-1
Minimize the size	0	0	-1	-1	0
Mobile	1	1	-1	-1	0
Prevents air bubbles	0	0	0	0	0
Prevents disease transmission	0	0	-1	0	0
Transfuse various types of blood (PRCB, FFB, platelets, etc.)	1	1	0	-1	1
Σ+	+7	+7	+6	+6	+6
Σ-	-5	-4	-9	-10	-7
Σ	**2**	**3**	**-3**	**-4**	**-1**

Appendix A-6: Gantt Chart

ID		Task Name	Duration	Start	Finish	Predecessors	Resource Names	Jan 25, '09		Feb 1, '09
								T F S S M T W T F S		S M T W
1		Interview Sponsor	2 days	Fri 1/23/09	Mon 1/26/09		Ashley	Ashley		
2		Interview Professionals	1 day?	Fri 1/30/09	Fri 1/30/09		Contacts,phone,filed t			Contacts,phone,filed t
3		QFD	1 day?	Tue 1/27/09	Tue 1/27/09		Dan-All	Dan-All		
4		DR1 Outline	4 days	Tue 1/27/09	Fri 1/30/09					
5		DR1 Practice	1 day?	Wed 1/28/09	Wed 1/28/09					
6		DR1 Paper-Final Revision	1 day	Wed 1/28/09	Wed 1/28/09					
7		Design Review 1	1 day	Thu 1/29/09	Thu 1/29/09					
8		Professional Contacts	7 days	Sat 1/31/09	Mon 2/9/09	4	Dan,Devon			
9		Brain Storm	3 days	Sat 1/31/09	Tue 2/3/09		Brad-All			Brad-
10		Contact Sponsor	1 day	Wed 2/4/09	Wed 2/4/09		CAEN-Devon-All			
11		Design 1 (CAD)	8 days	Tue 2/3/09	Thu 2/12/09					
12		DR2 Outline	1 day	Mon 2/16/09	Mon 2/16/09		Ashley			
13		DR2 Revision	3 days	Mon 2/16/09	Wed 2/18/09					
14		DR2 Practice	2 days	Tue 2/17/09	Wed 2/18/09					
15		DR2 Final Paper Review	1 day	Wed 2/18/09	Wed 2/18/09					
16		DR2 Presentation	1 day	Thu 2/19/09	Thu 2/19/09					
17		DR2 Final Paper Due	1 day	Fri 2/20/09	Fri 2/20/09					
18		Order Parts	5 days	Mon 2/16/09	Fri 2/20/09		Dan-All			
19		Safety Review	1 day	Mon 2/16/09	Mon 2/16/09					
20		Spring Break	10 days	Fri 2/20/09	Thu 3/5/09		SB			
21		Manufacturing Plan Alpha-Alpha	9 days	Wed 3/4/09	Mon 3/16/09		Brad-All			
22		Manufacture	2 days	Wed 3/18/09	Thu 3/19/09		Devon-All			
23		Contact Sponsor	1 day	Fri 3/20/09	Fri 3/20/09	22	Ashley-All			
24		Test	2 days	Mon 3/23/09	Tue 3/24/09	23	Dan-All			
25		Final Design-CADDED	2 days	Mon 3/23/09	Tue 3/24/09		Brad-All			
26		DR3 Presentation Outline	1 day	Fri 3/20/09	Fri 3/20/09					
27		DR3 Practice/Final Revision	1 day	Fri 3/20/09	Fri 3/20/09					
28		DR3 Presentation	1 day	Mon 3/23/09	Mon 3/23/09					
29		New Manufacturing Plan-Alpha	2 days	Wed 3/25/09	Thu 3/26/09		Devon-All			
30		Re-manufacture-Alpha Prototype	2 days	Fri 3/27/09	Mon 3/30/09		Ashley-All			
31		Test	2 days	Mon 3/30/09	Tue 3/31/09		Dan-All			
32		Final Report outlined and Assig	2 days	Fri 3/27/09	Mon 3/30/09		Brad-All			
33		Rough Copies of report sections	2 days	Wed 4/1/09	Thu 4/2/09					
34		DR4 tasking and practicing	2 days	Sun 4/5/09	Mon 4/6/09					
35		DR4	1 day	Tue 4/7/09	Tue 4/7/09					
36		Design Expo Prep	3 days	Wed 4/8/09	Fri 4/10/09		Devon-All			
37		Design Expo	1 day	Thu 4/16/09	Thu 4/16/09					
38		Final Report Revision	2 days	Thu 4/16/09	Fri 4/17/09					
39		Final Report Finalized	2 days	Mon 4/20/09	Tue 4/21/09					

Project: ghana gant	Task		Milestone ◆	External Tasks
Date: Tue 4/21/09	Split		Summary	External Milestone ◇
	Progress		Project Summary	Deadline ⇩

Page 1

ID		Task Name	Duration	Start	Finish	Predecessors	Resource Names	T	F	S	S	M	T	W	T	F	S	S	M	T	W
											Jan 25, '09							Feb 1, '09			
40		Final Report Due	1 day	Tue 4/21/09	Tue 4/21/09																
41		Party	7 days	Tue 4/21/09	Wed 4/29/09																

Project: ghana gant
Date: Tue 4/21/09

Task		Milestone	◆	External Tasks	
Split	....	Summary		External Milestone	◇
Progress		Project Summary		Deadline	⇩

Page 2

Dan,Devon

ad-All

CAEN-Devon-All

Ashley

Dan-All

SB

Brad-All

Project: ghana gant
Date: Tue 4/21/09

Task		Milestone	◆	External Tasks	
Split	....	Summary		External Milestone	◇
Progress		Project Summary		Deadline	⇩

Page 3

Project: ghana gant
Date: Tue 4/21/09

Task	▬▬▬▬	Milestone	◆	External Tasks
Split	··················	Summary	▼▬▬▬▼	External Milestone ◇
Progress	▬▬▬▬	Project Summary	▽▬▬▽	Deadline ⇩

Devon-All

Ashley-All

Dan-All

Brad-All

Devon-All

Ashley-All

Dan-All

Brad-All

Devon-All

Task	Milestone ◆	External Tasks
Split ··············	Summary	External Milestone ◇
Progress	Project Summary	Deadline ⇩

		Mar 22, '09							Mar 29, '09							Apr 5, '09							Apr 12, '09							Apr 19, '09							Apr 26, '09					
F	S	S	M	T	W	T	F	S	S	M	T	W	T	F	S	S	M	T	W	T	F	S	S	M	T	W	T	F	S	S	M	T	W	T	F	S	S	M	T	W	T	F

Task	▬▬▬▬
Split	····················
Progress	▬▬▬▬
Milestone	◆
Summary	▬▬▬
Project Summary	▭▭▭
External Tasks	▭▭▭▭
External Milestone	◇
Deadline	⇩

Project: ghana gant
Date: Tue 4/21/09

Appendix B: Bill of Materials

	Item	Quantity	Source	Catalog Number	Cost (USD)	Contact	Notes
Electrical Components	Printed Circuit Board 2-Layer 0.062"	1	Advanced Circuits	152	33.00	4pcb.com	
	32-Bit Microcontroller IC Input Max 6V Output Max 3.3V	1	Microchip	98M1936	7.50	newark.com	
	Voltage Regulator IC Voltage Regulator IC	1	Vishay Siliconix	57J5811	0.28	newark.com	
	Thick Film Chip Resistor 240ohm 125mW	1	Vishay Dale	59M6878	0.03	newark.com	
	Resistance: 4.7kohm Power Rating:2 50mW	14	Multicomp	01N7037	0.03	newark.com	
	Multilayer Ceramic Capacitor 100000pF	5	Multicomp	38K9284	0.02	newark.com	
	Ceramic Multilayer Capacitor 1µF	1	Taiyo Yuden	86H5309	0.03	newark.com	
	Ceramic Multilayer Capacitor 2.2µF	2	Taiyo Yuden	08P5127	0.04	newark.com	
	Ceramic Multilayer Capacitor 10µF	1	Taiyo Yuden	08P5151	0.05	newark.com	
	Ceramic Multilayer Capacitor 22µF	1	Taiyo Yuden	18J2704	0.17	newark.com	
	Bipolar Transistor Vceo: 40V	1	ON Semiconductor	83H7337	0.03	newark.com	
	Thick Film Chip Resistor 100ohm 125mW	2	Vishay Dale	59M6851	0.03	newark.com	
	Voltage Regulator IC Input Max 35V Output Max 5.1V	1	STMicroelectronics	89K0781	0.42	newark.com	
	Diode Vrrm: 50V	1	SPC Multicomp	14J9392	0.07	newark.com	
	Memory Card Headers MicroSD	1	3M Electronics	517-2908-05WB-MG	1.19	mouser.com	
	DC Power Jack PCB 2.1mm	1	Kobiconn	16PJ031	1.24	mouser.com	
	AC Adaptor 12VDC/1A	1	Xicon	412-112104	13.38	mouser.com	
	Rotary Encoders 12mm	3	Alpha (Taiwan)	318-ENC130175F-12PS	2.26	mouser.com	
	USB Connector	1	Molex	538-67503-1020	1.12	mouser.com	
	Programmable MEMS Oscillator 4MHz	1	SiTime	788-8002AI233E-4.0T	1.83	mouser.com	
	LCD Character Display Module	1	Hantronix	HDM16216L-5-E30S	9.18	mouser.com	
	Diode Schottky 1A 40V	1	Vishay	MSS1P4-E3/89AGICT-ND	0.07	digikey.com	
	LED Red/Green Clear	1	CML Innovative Technologies	CMD15-22SRUGC	0.27	digikey.com	
	Buzzer 2.048kHz	1	CUI Inc.	CEM-1203(42	0.72	digikey.com	
	Motor Driver IC	1	Toshiba	TB6612FNG	0.77	digikey.com	
Mechanical Components	5.90" Stroke 107 lb. 12 VDC Linear Actuator	1	Surplus Center	5-1577-6	79.95	surpluscenter.com	
	Linear Actuator Mounting Bracket Light Duty	1	Surplus Center	5-1577-B	8.95	surpluscenter.com	
	3/4" Extension Cabinet Slide with Lift-Out Release	1	McMaster-Carr	11965A131	17.82	mcmaster.com	
	White 8" Zip Tie	5	Gardner Bender	100046280	2.95	homedepot.com	
	1/2 In. x 4 Ft. x 8 Ft. Plywood Board	1	Millstead	787792	6.77	homedepot.com	
	30" x 36" 0.220" Acrylic Sheet (Plexiglass)	1	Optix	121015	N/A	homedepot.com	
	3/4 In. x 6 Ft Foam Pipe Insulation	1	Armacell	420390	7.49	homedepot.com	
	1/4" Dia. Screw 2" Length	2	Various	Various	0.50	Carpenter Brothers Hardware	
	1/4" Nut	3	Various	Various	0.40	Carpenter Brothers Hardware	
	3/16" Dia. Screw 1" Length	2	Various	Various	0.50	Carpenter Brothers Hardware	
	3/16" Wing Nut	2	Various	Various	0.40	Carpenter Brothers Hardware	
	1/4" Dia. Bolt 3" Length	1	Various	Various	0.75	Carpenter Brothers Hardware	
Medical Components	1/8" PP Liquid/Gas Check Valves	6	United States Plastic Corp.	64046	1.10	usplastic.com	
	Y Connector - 2.5mm ID	3	DirectMed	CY-001	N/A	directmed.com	
	20cc Syringe, Luer Lock Tip	2	Kendall	8881520657	3.18	medline.com	
	Silcone Tubing (Biosil 1250) 50' Inner Diameter: 5/32"	1	Saint-Gobain Performance Plastics	31005	N/A	plastics.saint-gobain.com	
	Medical IV Bag 2000mL	3	DirectMed	Unknown	N/A	directmed.com	
Processes	3D Printing/Rapid Prototyping (ABS Plastic)	1	UM3D Lab	Dimension FDM Elite 3D Printer	54.00	um3d.dc.umich.edu	Rapid Prototyped Slider Component
Miscellaneous	White Spray Paint (Plastic, Wood, Metal Compatible)	1	Krylon	2420 or 2320	5.79	Carpenter Brothers Hardware	

Appendix C: Description of Engineering Changes since Design Review III

The final prototype of the automatic exchange transfusion device exhibited a handful of engineering changes since Design Review #3. Most of these changes were discovered to be necessary during the manufacturing process where components were physically coming together and issues became more evident. Outlined below are the changes that were made with descriptions why. The subsequent pages consist of Engineering Change Notices (ECN) which provides more detailed and dimensioned descriptions of our design changes.

Syringe Holding Bracket
The holding piece that presses the syringes down onto the holding block went through a few design changes after Design Review #3. After we realized during the assembly process that its dimensions did not allow a normal force to hold the syringes down, we had to make the bracket legs less thick. However, instead of just reducing the thickness of the legs, we installed a foam layer at the bottom of the legs to allow the bracket to comply with various syringe diameters, rather than just having a rigid bracket bolted down on top. We also eliminated fillets throughout the bracket to make manufacturing easier. Also, since all of the component will be placed in to a protective case, 90˚ edges would not be expected to cause any safety issues with the user.

Syringe Holding Base
The base that the syringes lie upon had a design change regarding the slot that was placed in the top face. The slot was originally designed into the top face to accept the syringe collar and prevent it from sliding in the direction of the slider during operation. The slot was originally 0.1 in. and it was determined that it would be easier from a manufacturing standpoint to fabricate a piece that simply screws onto the base to hold the collars in.

Syringe Collar Holding Plate
As mentioned in the preceding paragraph, a design change to a moveable plate, rather than a slot was issued in order to hold the syringe collars to the base. This consisted of a simple piece of plexiglass dimensioned to the geometry of the block, with two 3/16" screws and wing nuts to hold it in place. The engineering change notice regarding this change provides an illustrative explanation of this modification.

Slider Holding Gap
An issue arose during initial testing regarding the plunger head and the gap of the slider that it is placed in. When the linear actuator is extending the plunger head was in contact with the far face of the gap. However, when the linear actuator changed directions from extension to retraction, there was a period of play where the plunger head had to change contact faces in the gap of the slider. This play accounted for a considerable distance of actuator movement, and during the course of a whole transfusion, the displacement losses would add up. Therefore, we had to redesign the slider gap to fit the plunger head more snugly and avoid the region of play between extension and retraction. To solve the problem we installed a 1/16" layer of foam to the far side of the slider to reduce the size of the gap. This solution solved the issue of play during linear actuator direction changes.

Protective Casing
The last engineering change we conducted after Design Review #3 was a material change for the protective casing which would surround of all of the system components. Instead of purchasing or fabricating a rigid one-piece protective casing, we designed one that is made of transparent plexiglass consisting of four interlocking walls and an opening top via hinges. Plexiglass was chosen because it can protect the system components, while still allowing observation of the system. The four walls were designed with interlocking corners, which were glued together with super glue.

Engineering Change Notices
Please refer to the following engineering change notices (ECN) for illustrative descriptions of these changes.
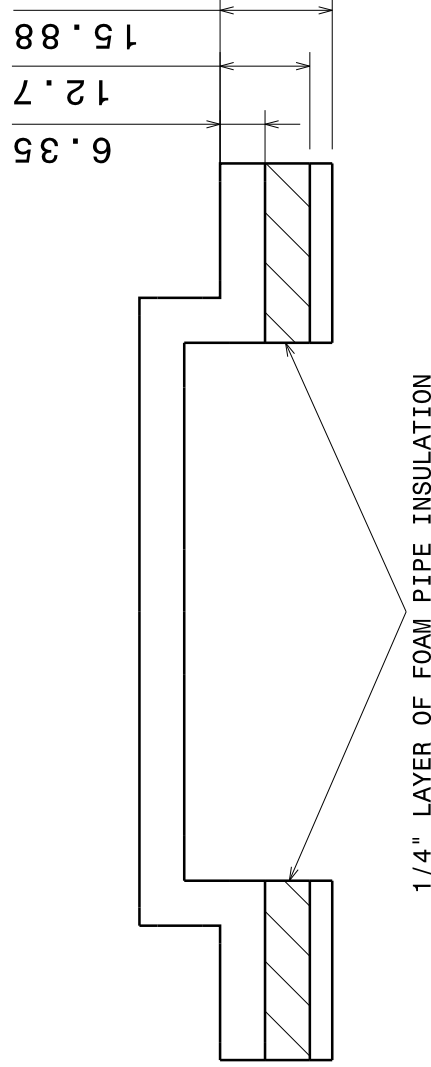
ENGINEERING CHANGE NOTICE

WAS:

R2.54

R2.54

R2.54

R6.35

6.35

IS:

6.35

12.7

15.88

1/4" LAYER OF FOAM PIPE INSULATION
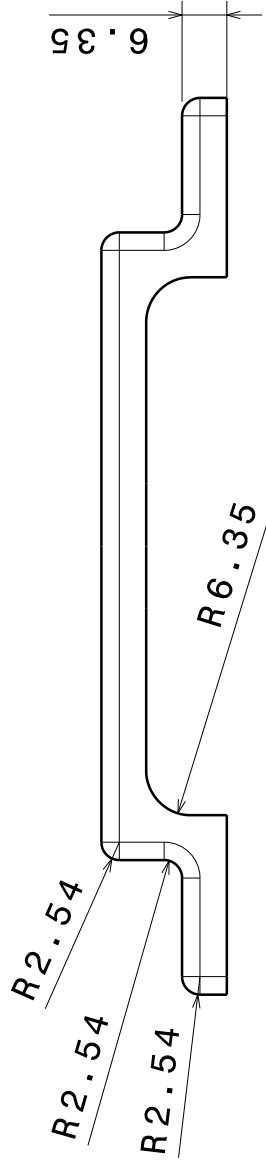
NOTES:
ALL FILLETS ELIMINATED
TO AID MANUFACTURABILITY

LAYER OF PIPE INSULATION ADDED
TO ALLOW HOLDING BRACKET A
TIGHTER AND MORE CUSTOM FIT
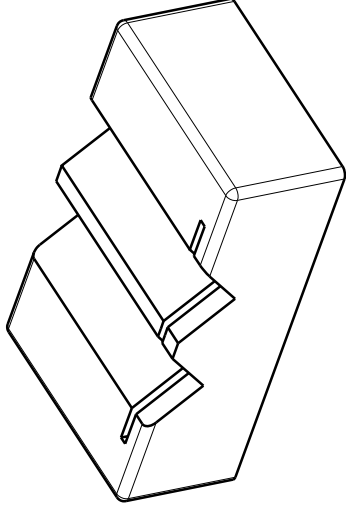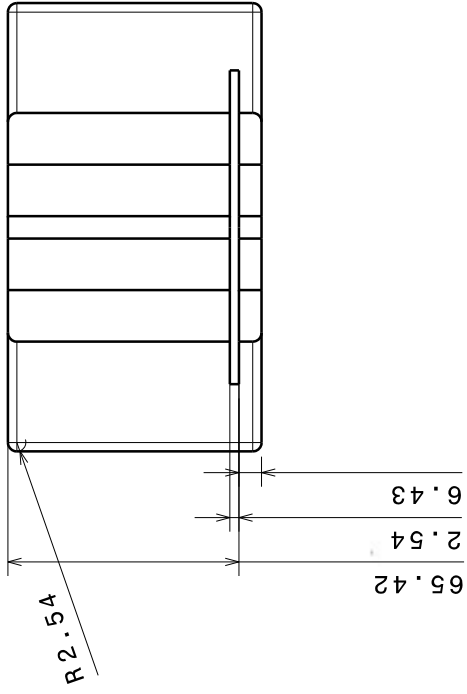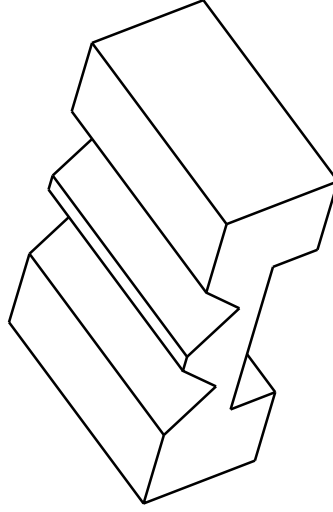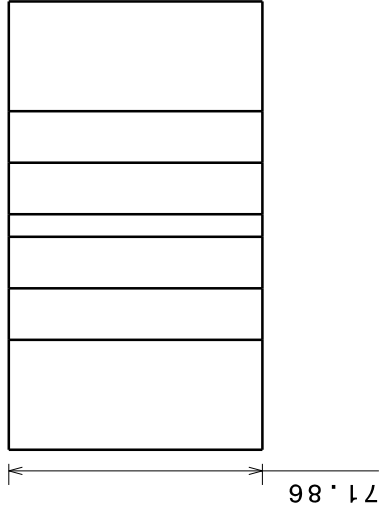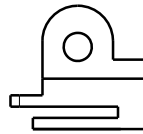ON SYRINGES

ALL DIMENSIONS IN MM
UNLESS OTHERWISE NOTED

UNIVERSITY OF MICHIGAN COLLEGE OF ENGINEERING		
PROJECT: AUTOMATED EXCHANGE TRANSFUSION DEVICE		
SYRINGE HOLDING BRACKET		
ENGINEER: DAN CIKO		4/20/2009
PROJECT MANAGER: ASHLEY MAY		4/20/2009
SPONSOR: KATHLEEN SIENKO		4/20/2009

ENGINEERING CHANGE NOTICE

WAS:

R2.54

65.42
2.54
6.43

IS:

71.86

15.24

64.77

NOTES:
ELIMINATED FILLETS
TO AID MANUFACTURABILITY

ELIMINATED COLLAR SLOT
TO AID MANUFACTURABILITY

REMOVED RECTANGLE OF MATERIAL
TO MAKE ROOM FOR TRACK
AND REDUCE WEIGHT

ALL DIMENSIONS IN MM

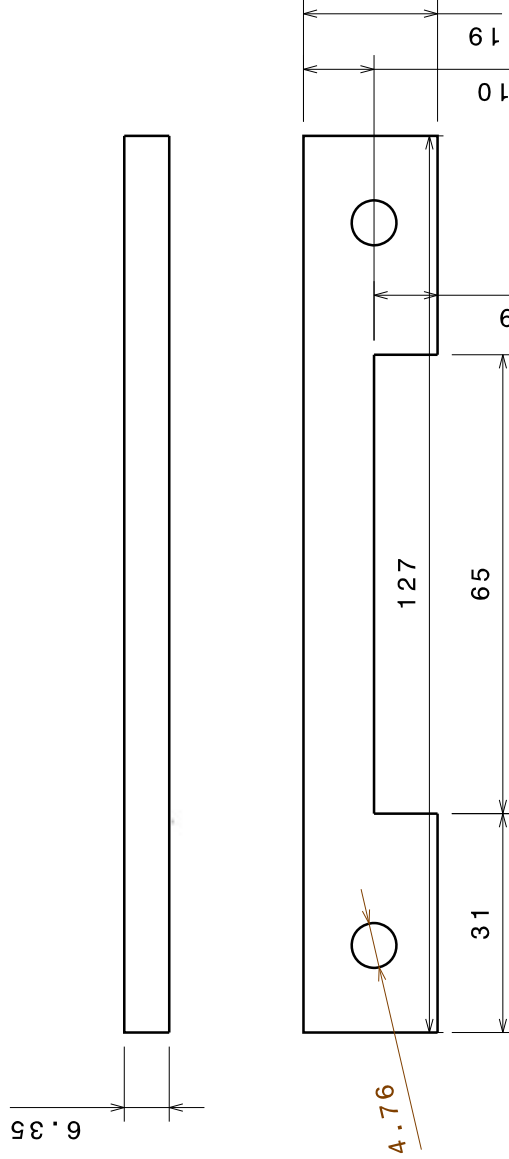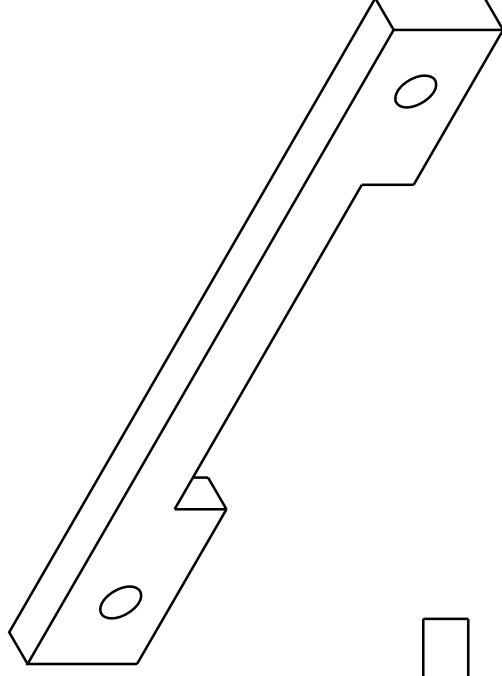| UNIVERSITY OF MICHIGAN COLLEGE OF ENGINEERING	
PROJECT: AUTOMATED EXCHANGE TRANSFUSION DEVICE	
SYRINGE HOLDING BASE	
ENGINEER: DAN CIKO	4/20/2009
PROJECT MANAGER: ASHLEY MAY	4/20/2009
SPONSOR: KATHLEEN SIENKO	4/20/2009
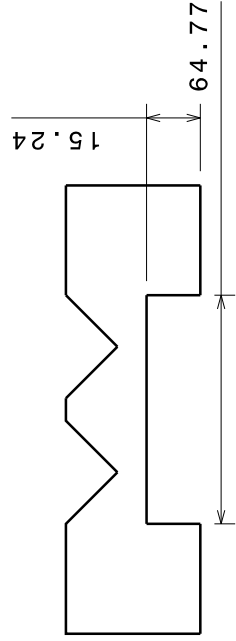
ENGINEERING CHANGE NOTICE
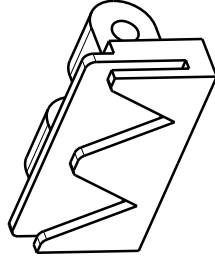
NOTES:
ADDED 1/16" LAYER OF FOAM PIPE INSULATION
TO REDUCE THE AMOUNT OF PLAY
OF THE PLUNGER HEAD IN THE SLOT

WAS:
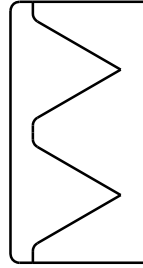
IS:

1/16" LAYER OF FOAM PIPE INSULATION
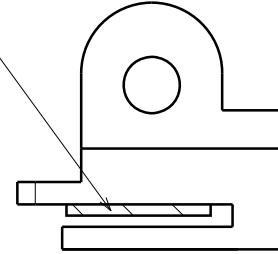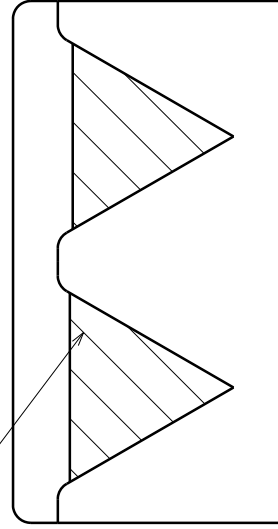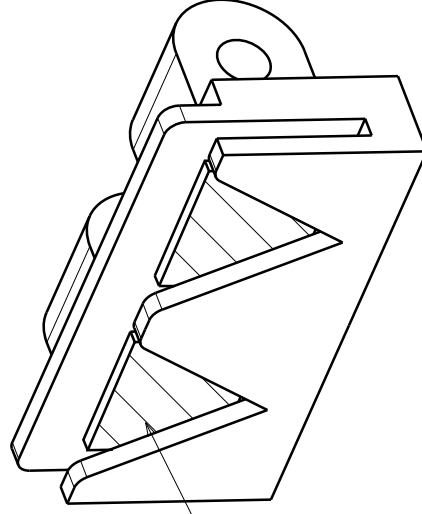
UNIVERSITY OF MICHIGAN COLLEGE OF ENGINEERING	
PROJECT: AUTOMATED EXCHANGE TRANSFUSION DEVICE	
SLIDER	
ENGINEER: DAN CIKO	4/20/2009
PROJECT MANAGER: ASHLEY MAY	4/20/2009
SPONSOR: KATHLEEN SIENKO	4/20/2009

ENGINEERING CHANGE NOTICE

NOTES:
MATERIAL IS 0.22" PLEXIGLASS

HOLES ACCOMODATE 3/16" SCREWS

19

10

6

127

65

31

6.35

4.76

ENGINEERING CHANGE NOTICE
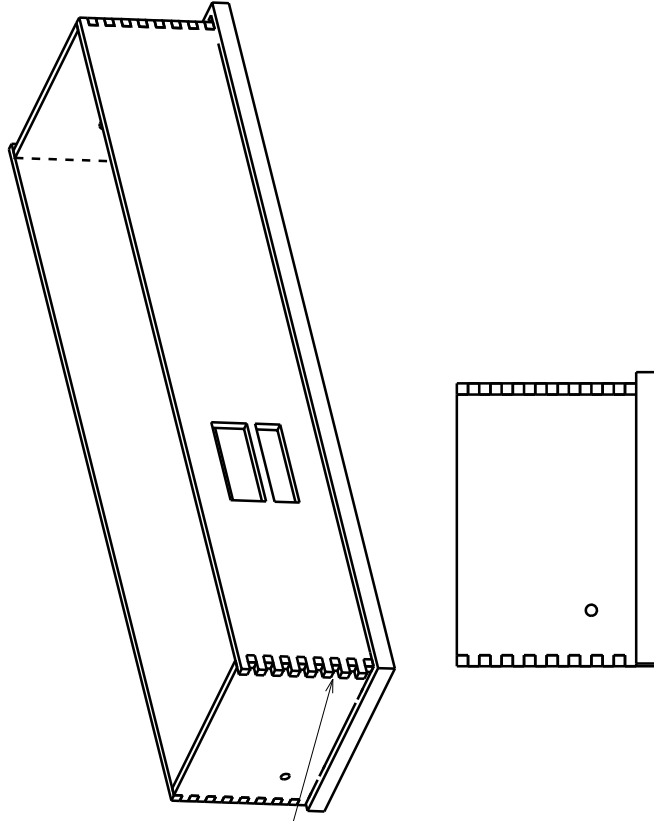
INTERLOCKING CORNERS

NOTES:
MATERIAL IS 0.22" PLEXIGLASS

UNIVERSITY OF MICHIGAN COLLEGE OF ENGINEERING		
PROJECT: AUTOMATED EXCHANGE TRANSFUSION DEVICE		
PROTECTIVE CASING		
ENGINEER: DAN CIKO		4/20/2009
PROJECT MANAGER: ASHLEY MAY		4/20/2009
SPONSOR: KATHLEEN SIENKO		4/20/2009

HOLE FOR POWER CORD

R3.18

6.35

6.35

SLOTS FOR USER INTERFACE
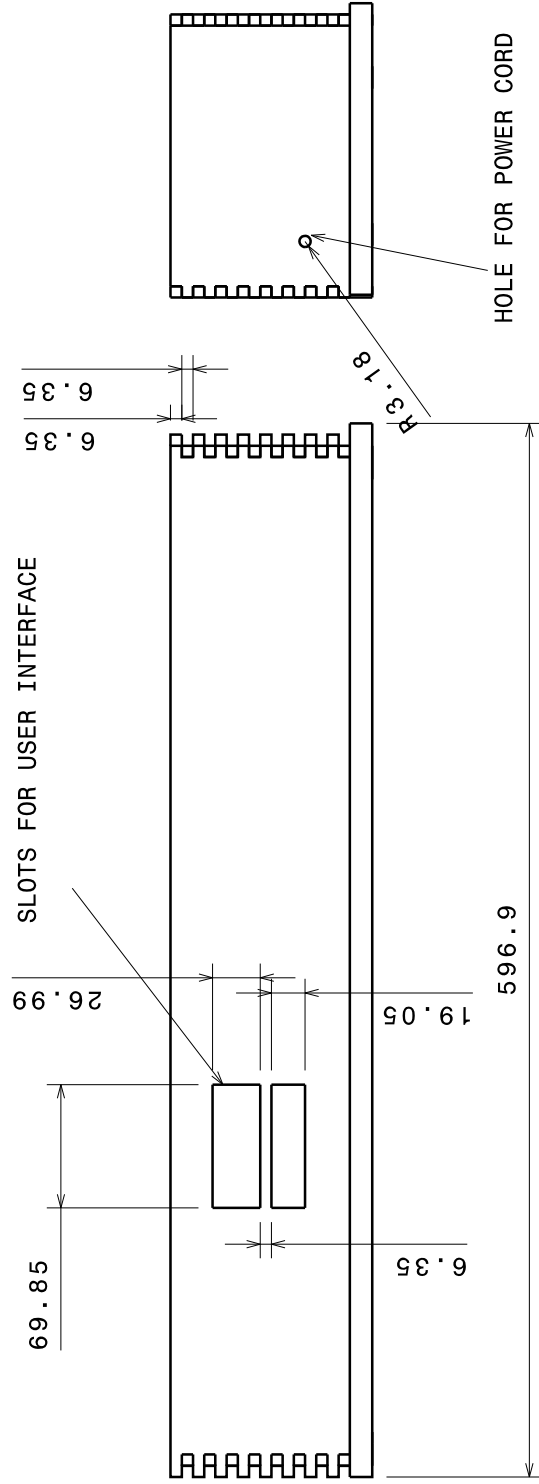
596.9

26.99

19.05

69.85

6.35

SLOT FOR SYRINGES
TO PROTRUDE FROM

38.1

76.2

165.1

UNIVERSITY OF MICHIGAN COLLEGE OF ENGINEERING		
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Appendix D: Design Analysis

Appendix D-1: Material Selection (Functional Performance)

The two main components of the automated ET device that will require manufacturing in Ghana are the slider and the syringe holder. Due to the minimal amount of available resources in Ghana, an appropriate material must be selected, which will be readily available in Ghana, while still adhering to the appropriate engineering standards. The Cambridge Engineering Selector (CES) was utilized to determine a variety of materials applicable for the necessary engineering applications. The parameter analysis of the syringe holder and the slider was used to determine material constraints.

Slider

The function of the slider is to provide a holding mechanism for the plunger of the syringe; therefore, the plunger of the syringe will extend and retract depending on the position of the linear actuator compared to the position of the syringe holder. The slider may experience bending from the force of the syringe; therefore, the syringe holder must have a yield strength of at least 1.06 MPa (the in depth calculation is provided in the "Slider" section of the Parameter Analysis). The slider is also prone to failure due to fracture. It was determined through a thorough engineering analysis that the slider should have a minimum fracture toughness of $0.0670\ MPa\sqrt{m}$. The slider must also have a maximum density of 2637 kg/m^3 to prevent the device from being over the weight specification. Not only are there various material property specifications, which need to be considered, but the material cost also needs to be investigated. The automated ET device must be as low cost as possible; therefore, a maximum cost of 2 USD/lb was determined.

The CES software was utilized to find an appropriate material, which adheres to all of the material specifications listed above. Various materials maintained the material requirements; therefore, a material index was determined. The function of the slider closely resembles that of a beam. One of the main objectives of the slider material is to minimize the weight, while constraining the prescribed strength; therefore, the material index expressed in Eq. 14 below was utilized for this application

$$\frac{\sigma_y^{2/3}}{\rho} \quad \text{(Eq. 14)}$$

where σ_y is the yield strength and ρ is the density. For the slider, the logarithmic material index used was 770 inches. The material selection was limited by comparing the yield strength and density exhibited in Figure 69 (p. 84).

After analyzing the various materials above the material index exhibited in Figure 69 (p. 84), the following five materials were identified as the most applicable: hard rubber, polyethylene (high density, low/medium molecular weight), polymethylmethacrylate (molding and extrusion), polypropylene (copolymer, UV stabilized), and poly vinyl chloride (rigid, molding and extrusion). Polypropylene and poly vinyl chloride are the two best materials primarily due to its durability in various fluids, its low cost, and its availability in Ghana.

Hard rubber

Hard rubber would be a sufficient material to manufacture the slider. Hard rubber has a density ranging from 0.039 lb/in^3 (1080 kg/m^3) to 0.0462 lb/in^3 (1279 kg/m^3), which is significantly below the maximum material density of 0.096 lb/in^3 (2637 kg/m^3). The yield strength of hard rubber is well above 0.154 ksi (1.06 MPa) for the yield strength of hard rubber ranges from 8.7 ksi (60.0 MPa) to 11.6 ksi (80.0 MPa), almost sixty-times greater than the required yield strength. The fracture toughness of hard rubber is 0.182 $ksi\sqrt{in}$ (0.20 $MPa\sqrt{m}$), which is over 3 times greater than the required value.
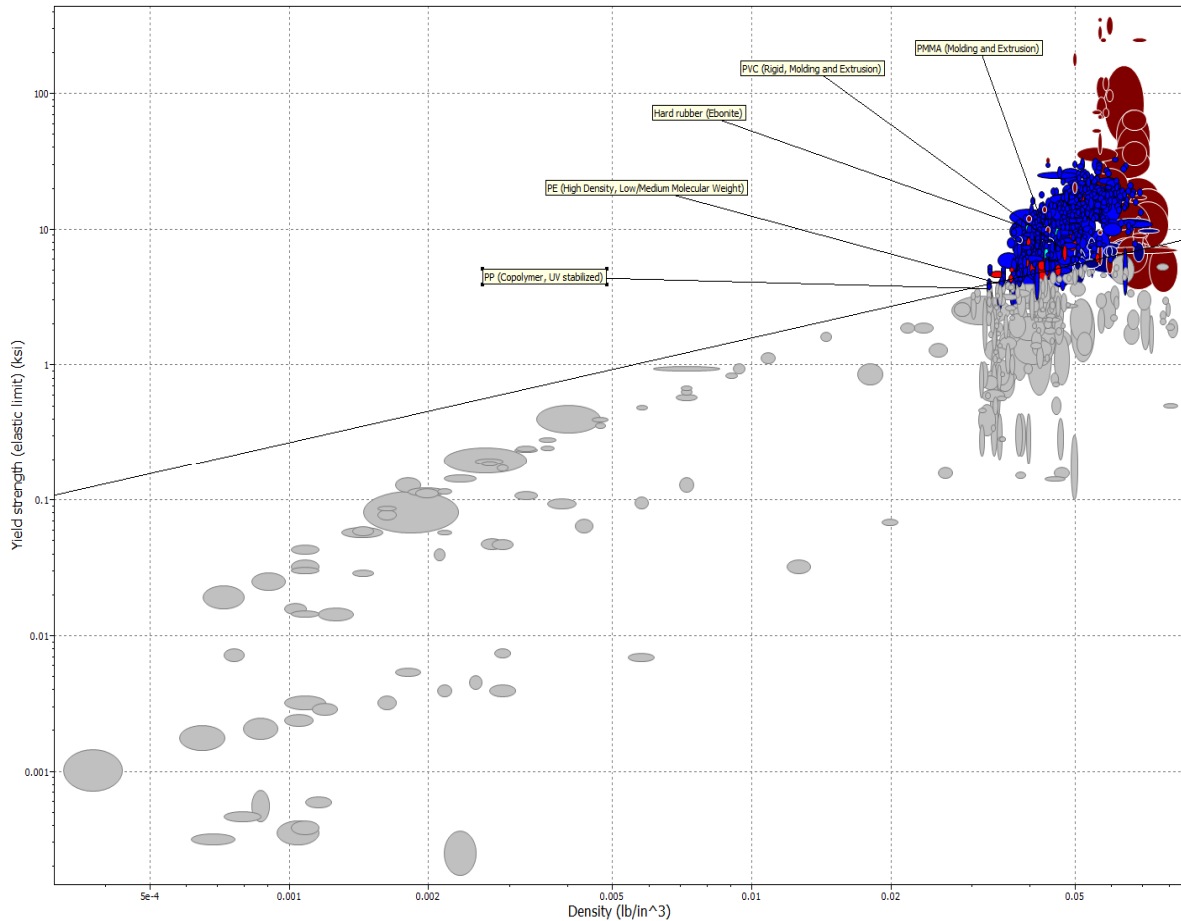
Figure 69. Slider material selection comparison using CES.

Hard rubber is resistive to most chemicals. It resists carboxylic acids and oils. It is also very resistive to fresh water, salt water, and alkalis; therefore, any leakage of blood should not affect the slider. The slider can also be easily cleaned with a bleach solution if created from hard rubber. It is also fairly resistive to acids, oils, and fuels. UV radiation should have a negligible impact on the material properties as well.

Hard rubber only absorbs 0.01 to 0.1 percent of water over a 24 hour period; therefore, the hard rubber should not become contaminated from any blood that may be spilt. Hard rubber has good adhesion to metal, which is necessary for the slider to be attached to the track. It is an opaque material. Hard rubber is fairly expensive compared to other materials. It costs between 1.30 and 1.43 USD/lb.

Polyethylene (High Density, Low/Medium Molecular Weight)
The slider could be produced out of polyethylene (*PE*). PE has a density ranging from 0.0342 lb/in^3 (947 kg/m^3) to 0.0345 lb/in^3 (955 kg/m^3), which is significantly below the maximum material density of 0.096 lb/in^3 (2637 kg/m^3). The yield strength of PE is almost twenty-times greater than the required yield strength of 0.154 ksi (1.06 MPa) for the yield strength of PE ranges from 2.8 ksi (19.3 MPa) to 3.9 ksi (26.9 MPa). The fracture toughness of PE ranges between 1.04 $ksi\sqrt{in}$ (1.14 $MPa\sqrt{m}$) and 1.9 $ksi\sqrt{in}$ (2.08 $MPa\sqrt{m}$), which is about twenty times greater than the required value.

If the slider is to be molded, the mold temperature is between 86 and 122 °F, while the molding pressure range is between 12 ksi (83 MPa) and 14.9 ksi (103 MPa). There is also a minimal amount of mold shrinkage for it only shrinks 1.5 to 4 percent of its original size.

PE is resistive to most chemicals. It is very resistive to fresh water, salt water, weak acids, and alkalis; therefore, blood and cleaning solutions (i.e. bleach) should not affect the material properties of the slider if made out of PE. It is fairly resistive to organic solvents. UV radiation should not have a significant impact on the material properties either.

PE absorbs between 0.005 and 0.01 percent of water over a 24 hour period; therefore, it should absorb a minimal amount of liquid if spilt on the slider. It is a translucent material; therefore, one can see through the material. PE is not too costly; it costs between 0.98 and 1.08 USD/lb.

Polymethylmethacrylate (Molding and Extrusion)
Polymethylmethacrylate (PMMA) would be an adequate material to produce the slider from. PMMA has a density ranging from 0.0423 lb/in^3 (1171 kg/m^3) to 0.0434 lb/in^3 (1201 kg/m^3), which is significantly below the maximum material density of 0.096 lb/in^3 (2637 kg/m^3). The yield strength of PMMA is over fifty-times greater than the required yield strength of 0.154 ksi (1.06 MPa) for the yield strength of PMMA ranges from 7.8 ksi (53.8 MPa) to 10.5 ksi (72.7 MPa). The fracture toughness of PMMA ranges between $0.637\ ksi\sqrt{in}$ ($0.670\ MPa\sqrt{m}$) and $1.46\ ksi\sqrt{in}$ ($1.60\ MPa\sqrt{m}$), which is about ten times greater than the required value.

If the slider is to be molded, the mold pressure range will between 4.99 ksi (34 MPa) and 20 ksi (138 MPa). There will only be 0.1 to 0.8 percent mold shrinkage from the original size, which is minimal.

PMMA resists most chemicals. It is very resistive to fresh water, salt water, and alkalis; therefore, bodily fluids will not affect the material properties of the slider. A base, such as bleach, will not affect the slider either. PMMA does not resist organic solvents very well. Acids will cause corrosion to occur. Sunlight will not impact the slider either.

PMMA absorbs between 0.1 and 0.4 percent of water over a 24 hour period. PMMA is not overly expensive for it costs a maximum of 1.22 USD/lb.

Polypropylene (Copolymer, UV Stabilized)
The slider could be manufactured from polypropylene (PP). PP has a density ranging from 0.0325 lb/in^3 (900 kg/m^3) to 0.0328 lb/in^3 (908 kg/m^3), which is significantly below the maximum material density of 0.096 lb/in^3 (2637 kg/m^3). The yield strength of PP is over twenty-times greater than the required yield strength of 0.154 ksi (1.06 MPa) for the yield strength of PP ranges from 3.42 ksi (23.6 MPa) to 3.91 ksi (27.0 MPa). The fracture toughness of PP ranges between $1.37\ ksi\sqrt{in}$ ($1.51\ MPa\sqrt{m}$) and $1.44\ ksi\sqrt{in}$ ($1.58\ MPa\sqrt{m}$), which is about twenty times greater than the required value.

If the slider is to be molded, the mold temperature of PP is between 451 and 497 °F, while the molding pressure range is between 0.394 ksi (2.72 MPa) and 1.22 ksi (8.41 MPa). The mold will also only shrink 1.41 to 1.51 percent of its original size.

PP resists most chemicals. It is very resistive to fresh water, salt water, acids, alkalis, and organic solvents; therefore, the slider will not corrode from any fluids. PP is also very durable to sunlight (UV radiation).

PP absorbs between 0.0195 and 0.0205 percent of water over a 24 hour period. It is a translucent material. It is slow burning. PP has a unit cost of 0.968 to 1.07 USD/lb, which is not too expensive. It is typically used for medical components.

Poly Vinyl Chloride (Rigid, Molding and Extrusion)
Poly Vinyl Chloride (PVC) would be a sufficient material to manufacture the slider from. PVC has a density ranging from 0.047 lb/in^3 (1301 kg/m^3) to 0.0538 lb/in^3 (1489 kg/m^3), which is significantly below the maximum material density of 0.096 lb/in^3 (2637 kg/m^3). The yield strength of PVC is almost forty-times greater than the required yield strength of 0.154 ksi (1.06 MPa) for the yield strength of PVC ranges from 6.00 ksi (41.4 MPa) to 7.64 ksi (52.7 MPa). The fracture toughness of PVC ranges between 3.3 $ksi\sqrt{in}$ (3.6 $MPa\sqrt{m}$) and 3.5 $ksi\sqrt{in}$ (3.8 $MPa\sqrt{m}$), which is about sixty times greater than the required value.

If the slider is to be molded, the mold temperature of PVC is between 351 and 390 °F, while the molding pressure range is between 9.98 ksi (68.8 MPa) and 39.9 ksi (275 MPa). The mold will also only shrink 0.2 to 0.6 percent of its original size.

PVC resists most chemicals. It is very resistive to fresh water, salt water, acids, and alkalis; therefore, the slider should be durable when exposed to such fluids. PVC is fairly resistive to organic solvents. PVC is also very durable to sunlight (UV radiation).

PVC absorbs between 0.04 and 0.40 percent of water over a 24 hour period. It is a transparent material. It is self-extinguishing. PVC has a unit cost of 0.689 to 0.758 USD/lb, which is one of the cheapest materials. PVC is also a readily available material in Ghana.

Syringe Holder
The function of the syringe holder is to provide a suitable housing unit for the syringes. The syringe holder will experience a minimal force exerted by the weight syringes; therefore, the weight of the syringes is considered negligible in determining the material properties of the syringe holder. The lip of the syringe will exert a force on the side of the syringe holder when the linear actuator is extending, or applying a force to the syringe. The force from the lip of the syringe should also not affect the material property greatly for the force from linear actuator will be minimal due to the low speed of the linear actuator. The stress exerted by the lip of the syringe onto the syringe holder will be approximately the same the stress exerted on the slider; therefore, it can be assumed that the maximum stress exerted on the syringe holder from the lip of the syringe is approximately 1.06 MPa. The syringe holder must also have a maximum density of 2637 kg/m^3 to prevent the device from being over the weight specification. Not only are there various material property specifications, which need to be considered, but the material cost also needs to be investigated. The automated ET device must be as low cost as possible; therefore, a maximum cost of 2 USD/lb was determined.

The CES software was utilized to find an appropriate material, which adheres to all of the material specifications listed above. Various materials maintained the material requirements; therefore, a material index was determined. The function of the slider closely resembles that of a beam. One of the main objectives of the slider material is to minimize the weight, while constraining the prescribed strength; therefore, the material index expressed in Eq. 14 above was utilized for this application. For the slider, the logarithmic material index used was 770 inches. The material selection was limited by comparing the yield strength and density exhibited in Figure 70 below.
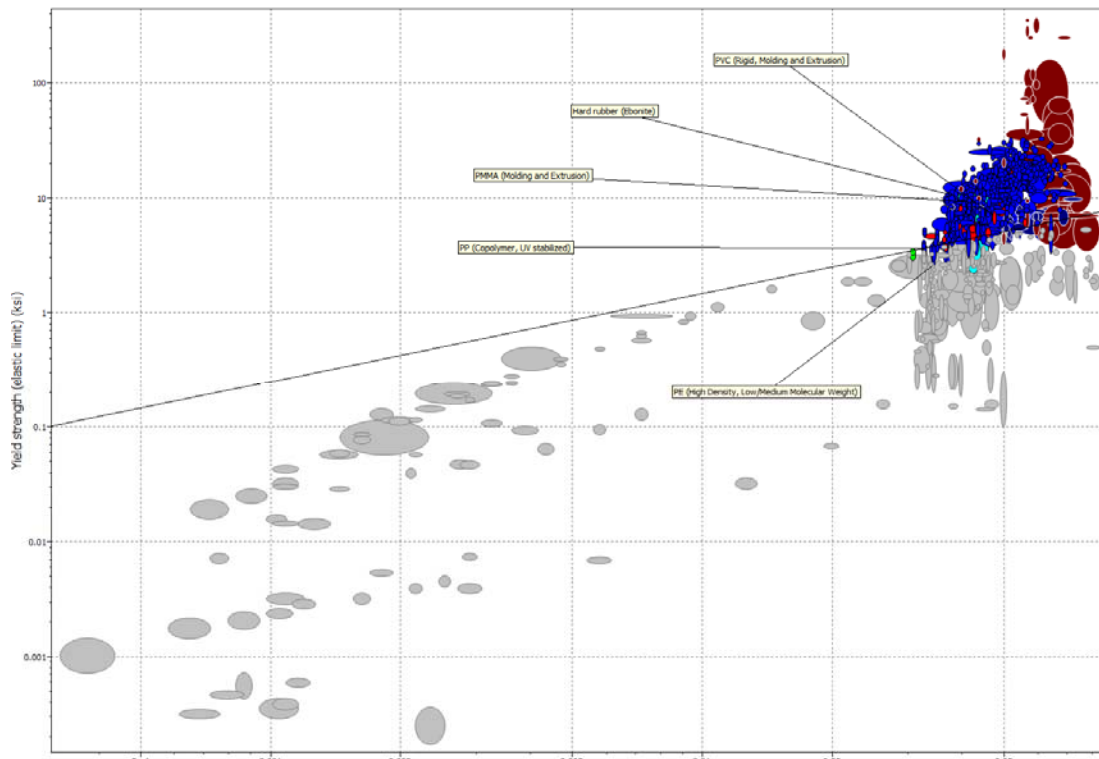
**Figure 70. Syringe holder material selection comparison using CES.**

Since the yield strength and the density remained the same, the material index also remained the same. The fracture toughness of the slider did have a large impact on the material selection; therefore the same five materials resulted from this analysis. Once again, polypropylene and poly vinyl chloride are the two best materials primarily due to its durability in various fluids, its low cost, and its availability in Ghana.

Appendix D-2: Material Selection (Environmental Performance)

The following two materials provide the best functional performance for the slider and the syringe holder: polypropylene (copolymer, UV stabilized) and poly vinyl chloride (rigid, molding and extrusion). Although PP and PVC are adequate materials for maintaining functionality of the system, the environmental performance of the two materials must be investigated before determining the best material for manufacturing.

The total mass of air emissions, water emissions, use of raw materials, and (solid) waste must be investigated for each material, which is depicted in Figure 1Figure 71 below. To determine the total emissions of each material, the amount of material must be specified by the mass of the material. The volume of the slider and the syringe holder totals to 2.51×10^{-4} m^3; therefore, the mass of PP would be approximately 0.373 kg (assuming a density of 1489 kg/m^3) and the mass of PVC would be approximately 0.228 kg (assuming a density of 908 kg/m^3).

Based off the SimaPro 7 software, PP generates 4% more air emissions than PVC; however, PVC generates more water emissions, uses more raw materials, and waste than PP. PVC emits 98% more water emissions than PP; however, PVC only releases 45.5 g of water emissions. PVC utilizes 37% more raw materials than PP. PVC also produces 69% more waste than PP, but PVC only gives off 120.4 g of waste.
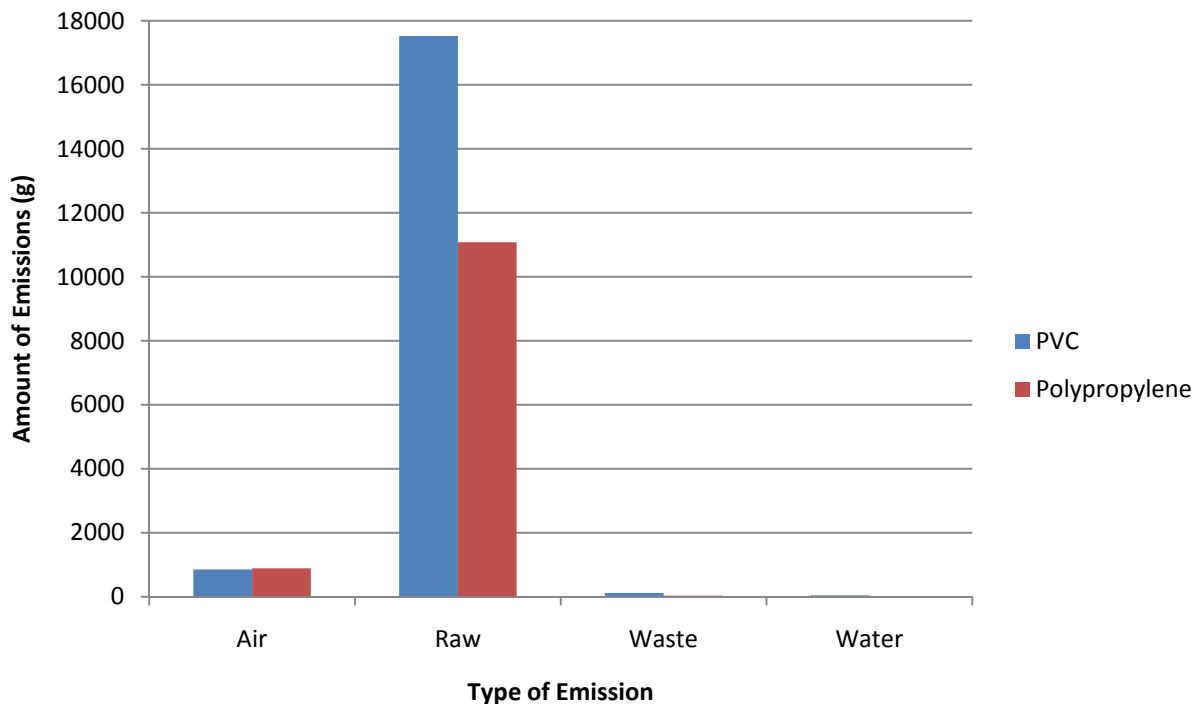

Figure 71. Total emissions generated by PVC and Polypropylene.

Each material can also have its environmental performance characterized based on the amount of carcinogen emissions, respiratory organics carbon consumption, respiratory inorganics carbon consumption, climate change, radiation, ozone layer destruction, ecotoxicity, acidification/eutrophication, and mineral depletion, which is depicted in Figure 72(p. 89). PVC emits more carcinogens, consumes more respiratory organic carbon, consumes more respiratory inorganic carbon, causes a greater amount of climate change, more ecotoxicity, more acidification/eutrophication, and more mineral depletion than PP.
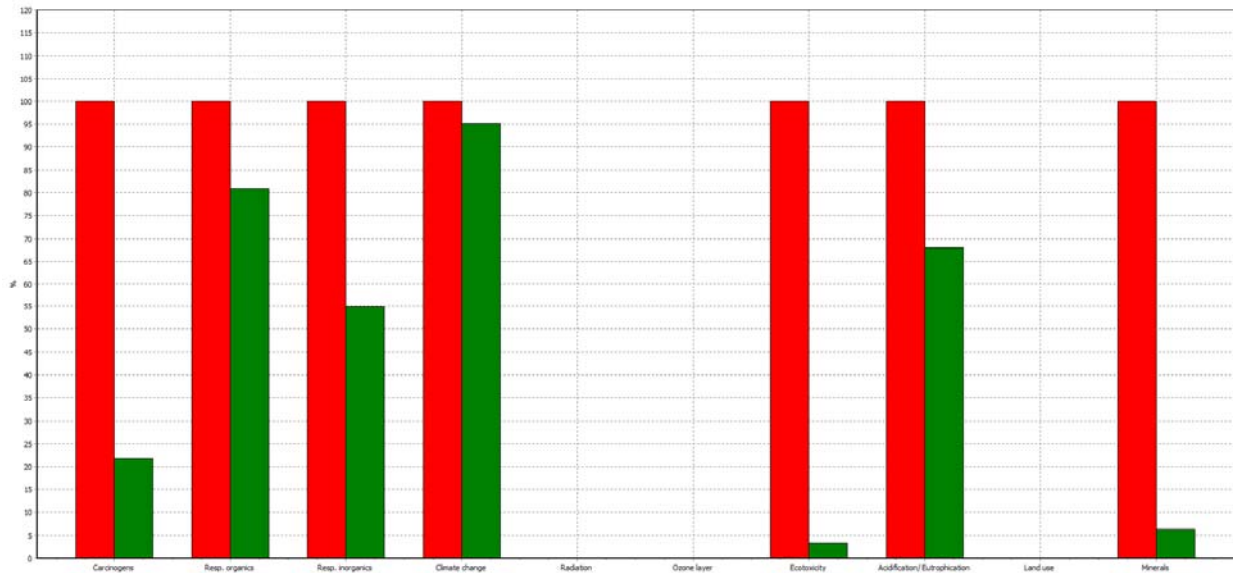
Figure 72. Environmental performance characterization comparing polypropylene and poly vinyl chloride.

PP and PVC also have an environmental impact on human health, ecosystem quality, and resources. Figure 73 below reveals the environmental damage on human health, ecosystem quality, and resources available. PVC has a greater environmental damage percentage than PP. A normalization graph of the environmental damage is represented in Figure 74 (p. 90). The normalization graph reveals that human health is impacted the most by the used of PP and/or PVC. The damage caused to human health is only 7 x 10^{-5}; therefore, the impact on human health is minuscule even though it makes up a majority of all environmental damage.
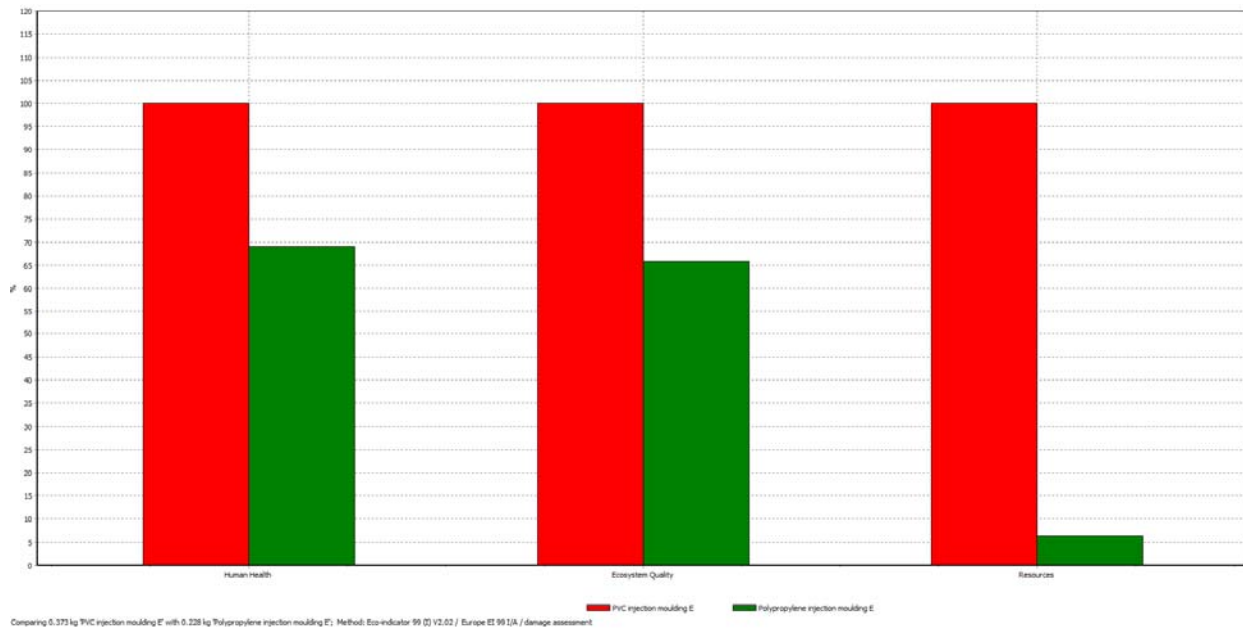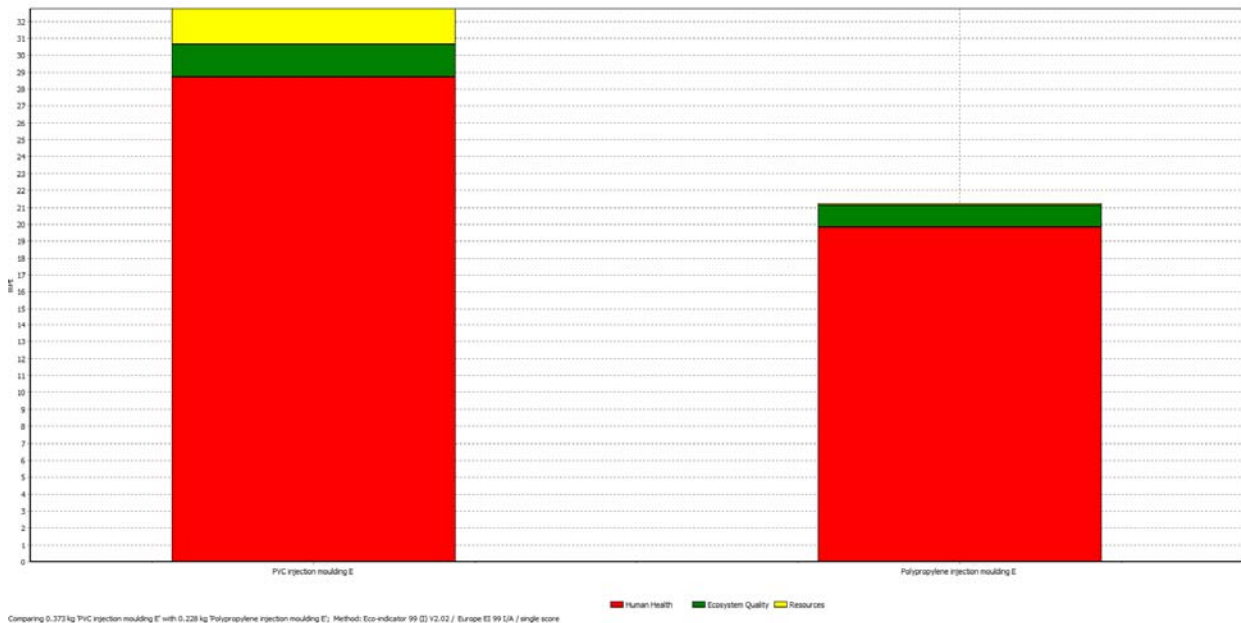

Figure 73. Environmental damage assessment of polypropylene and poly vinyl chloride.

89
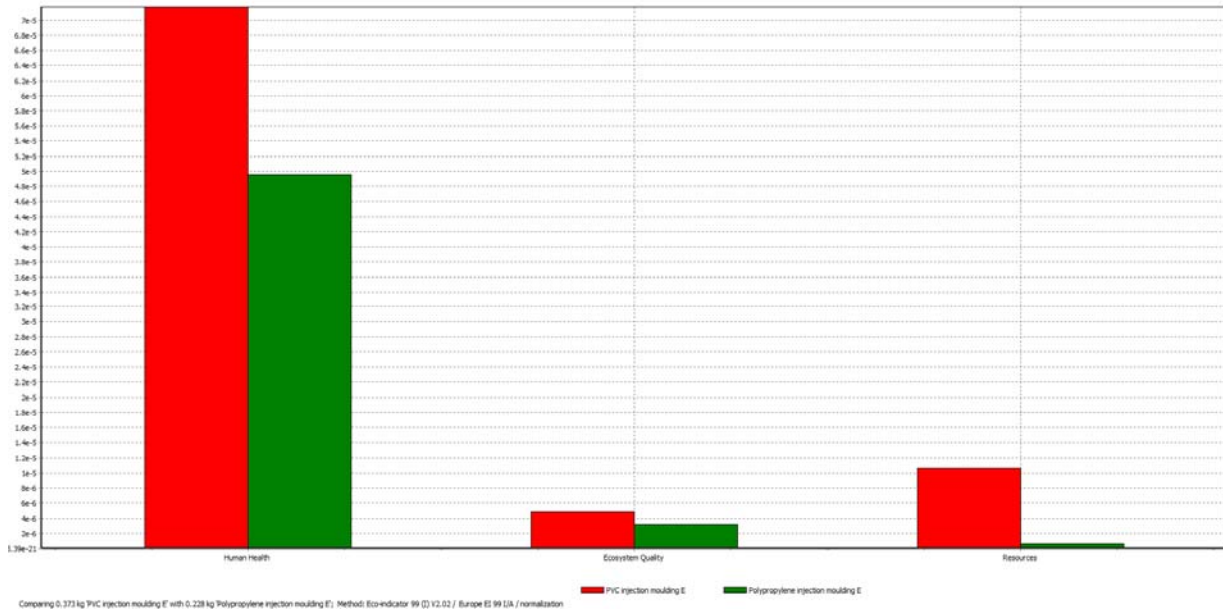
Figure 74. Normalized environmental damage assessment of polypropylene and poly vinyl chloride.

The environmental damage caused by PP and PVC can be compared through the use of single score comparison, which is depicted in Figure 75 below. Figure 75 reveals that PVC causes much more environmental damage than PP. Although PVC has a high EcoIndicator 99 point value of 33, the PVC should have a fairly long lifetime for it should not have to be replaced very frequently, if even at all.

Figure 75. Single score comparison of the environmental damage caused by polypropylene and poly vinyl chloride.

The PP seems to be the most environmentally friendly material; however, PP is not readily available in Ghana. Due to such circumstances, PVC would be the recommended material. PVC is also less expensive than PP.

Poly vinyl chloride (rigid, molding and extrusion) is also a transparent material; therefore, the material can also be used for the casing of the automated ET device. This will allow for medical personnel to view the operation of the device to insure that it is not malfunctioning. The casing will add a volume of 0.00254 m^3; therefore, causing the amount of PVC to increase to 4.16 kg. When the casing is produced from PVC as well, the EcoIndicator 99 point value increases to 480, which is depicted in Figure 76 below. As mentioned previously, the PVC components of the automated ET device should have a long lifecycle; therefore, the emissions will not be continuously emitted.
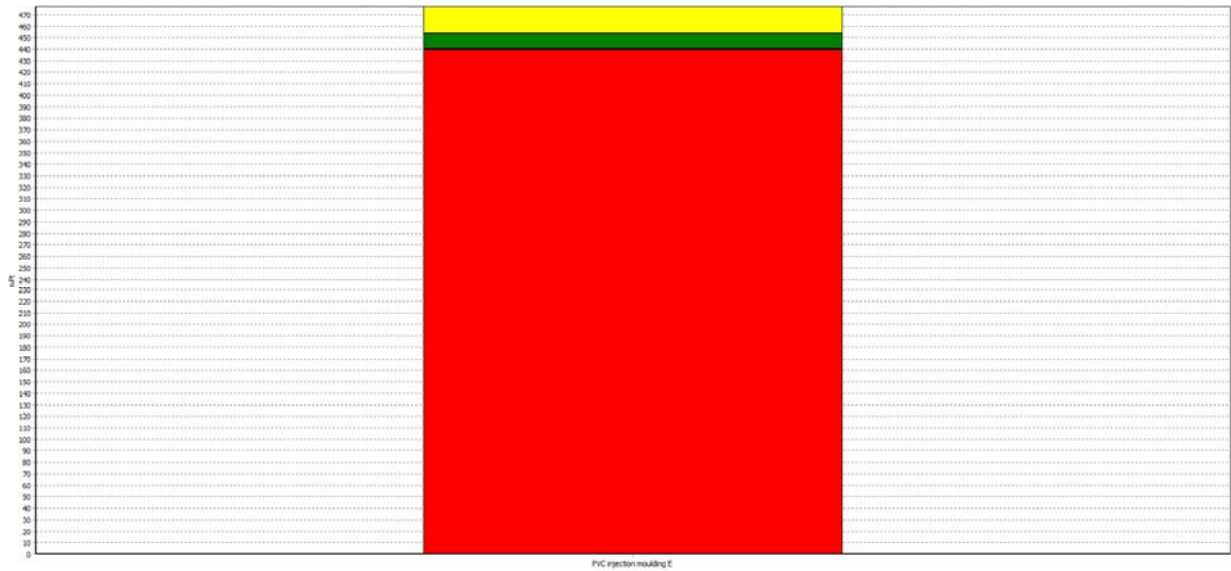
Figure 76. Single score of the environmental damage caused by poly vinyl chloride.

Appendix D-3: Manufacturing Process Selection

There may eventually be 1,000 automated ET devices around the world; however, all 1,000 automated ET devices will not be produced at the same time. The production volume for the automated ET device will consist of approximately 5 to 10. The automated ET device also consists of medical equipment, electrical components, and PVC components. A majority of the components, including the medical equipment and the electrical components, will be outsourced; therefore, only a small portion of the device will need to be produced.

Due to the minimal production volume, the manufacturing process should consist of the standard shop equipment (i.e. mill, band saw, hand drill, etc.). The standard shop equipment should also be readily available in Ghana; therefore, manufacturing can be done in-country. The manufacturing process should not extend far beyond the standard shop equipment for the tooling cost involved in processes, such as injection molding, are too expensive. The components are also not too intricate to require such complex processes. To make the manufacturing process slightly easier, a CNC mill could be programmed to manufacture the slider and the syringe holder, if accessible. The electrical components will need to be soldered onto the circuit board by hand; therefore, requiring access to a soldering iron. Other than those manufacturing the processes, the microprocessor will need to be programmed and the device will need to be assembled.

Appendix E: Electrical Components

Appendix E-1: Schematic of the Electronic Components

Schematic drawing — BloodTransfusion.SchDoc

OSC — SiT8002AI-23-33E-4.00000T — 4MHZ_3.3V — PIC32 XT1 — OE, Vdd, Power3, OSC_EN, GND, OUT, Ground

USB — USB TypeB Connector — USB_TYPEB — D+, D-, GND, 5V, USB D+, USB D-, Ground, USB 5V

REG3 — SI9183DT-33-T1-E3 - 3.3V — VIN, VOUT, GND, 3.3V/150mA, ^SD, BP — Power, Power3, C1 2.2uF, C2 2.2uF, C3 0.1uF, USB_5V

LCD — GDM1602K LCD — GND, VDD, V0, RS, R/W, E, DB0, DB1, LED-, LED+, DB7, DB6, DB5, DB4, DB3, DB2 — contrast adjustment - easy way — R17 1K, R18 7.5K, R16 7 — 7ohm resistor?? — DataRS, DataRW, DataE, Data0, Data1, Data7, Data6, Data5, Data4, Data3, Data2

PWR — POWERPLUG — POWER — Power12
D2 NRD4001 — **REG5 L7805CD2T** — IN, OUT, GND — Power — C9 22uF

MSS1P4 Schottky Diode — D1

PRG — ICSP_MINI_HEADER — MINI ICSP — Vihh, Power, Ground, ICSP_PGD, ICSP_PGC — PIC32_Vihh_Vihh, Power3, Ground, ICSP_PGD, ICSP_PGC

LED — CMD15-22SRUGC — R1 100 SOFT_LED, R2 100 SOFT_LED2 — Power3

BUZ — Buzzer CEM-1203 Buzzer — +, - — Power — BUZSIG R15 240 — T1 2N3904 — Ground

DRIVER — TB6612FNG — TB6612FNG Motor Driver Board — GND, VCC, AO1, AO2, BO2, BO1, MOT, GND, PWMA, AIN2, AIN1, ^STBY, BIN1, BIN2, PWMB, GND — Ground, Power3, Motor+, Motor-, Motor+, Power12, Ground, MotorPWM, MotorDir2, MotorDir1, Standby, MotorDir1, MotorDir2, MotorPWM, Ground

microSD — microSD slot, SPI mode — microSD_Slot — NC, DO, Vss, SCLK, Vdd, DI, ^CS, NC — SPI_DO_3, Ground, SPI_CLK_3, Power3, SPI_DI_3, SPI_CS_3, microSD_Slot

PIC32 — PIC32MX440F512H — PIC32MX440 — C8 10uF Tantalum

MOTOR — Header 2 — Motor+, Motor-
POS — Header 3 — Power3, Position, Ground — Position

SW1, SW2, SW3 — R6 4.7K, R7 4.7K, R8 4.7K, R9 4.7K, R10 4.7K, R11 4.7K, R12 4.7K, R13 4.7K, R14 4.7K — BTN1, BTN2, BTN3, ROTA1, ROTA2, ROTA3, ROTB1, ROTB2, ROTB3 — C10 1.0uF

C4 0.1uF, C5 0.1uF, C6 0.1uF, C7 0.1uF, R3 4.7K, R4 4.7K, R5 4.7K

Title		
Size A	Number	Revision
Date: 4/20/2009		
File: C:\Users\..\BloodTransfusion.SchDoc	Sheet of	Drawn By:

Appendix E-2: Microprocessor Specifications

Appendix E-3: Oscillator Specifications

SiTIME Programmable Oscillators

SiTime — It's about time™

This product is RoHS compliant.

SIT8002 SERIES: 3.3 VOLT, 2.5MM X 2.0MM PACKAGE

Features:
- One Time Programmable
- 1 MHz to 125 MHz Clock
- Ultra short leadtime
- Greater immunity from interference and ultra-reliable start up

PINOUT:
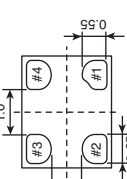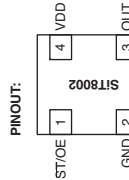Pad 1: Standby, Output Enable
Pad 2: Connect to Ground
Pad 3: Clock Output
Pad 4: Connect +3.3V
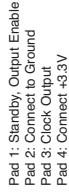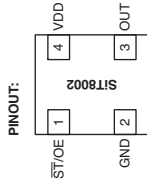
For quantities of 250 and up, call for quote.

MOUSER STOCK NO.	SiTime Part No.	Freq. (MHz)	Price Each 1	10	50	100
788-8002AI133E-1.8T	SIT8002AI-13-33E-1.84320T	1.84320	1.98	1.72	1.52	1.40
788-8002AI133E-2.04T	SIT8002AI-13-33E-2.04800T	2.04800	1.98	1.72	1.52	1.40
788-8002AI133E-3.6T	SIT8002AI-13-33E-3.68640T	3.68640	1.98	1.72	1.52	1.40
788-8002AI133E-4.0T	SIT8002AI-13-33E-4.00000T	4.00000	1.98	1.72	1.52	1.40
788-8002AI133E-5.0T	SIT8002AI-13-33E-5.00000T	5.00000	1.98	1.72	1.52	1.40
788-8002AI133E-6.0T	SIT8002AI-13-33E-6.00000T	6.00000	1.98	1.72	1.52	1.40
788-8002AI133E-8.0T	SIT8002AI-13-33E-8.00000T	8.00000	1.98	1.72	1.52	1.40
788-8002AI133E-10.0T	SIT8002AI-13-33E-10.00000T	10.00000	1.98	1.72	1.52	1.40
788-8002AI133E-12.0T	SIT8002AI-13-33E-12.00000T	12.00000	1.98	1.72	1.52	1.40
788-8002AI133E-13.0T	SIT8002AI-13-33E-13.00000T	13.00000	1.98	1.72	1.52	1.40
788-8002AI133E-13.5T	SIT8002AI-13-33E-13.56000T	13.56000	1.98	1.72	1.52	1.40
788-8002AI133E-14.3T	SIT8002AI-13-33E-14.31818T	14.31818	1.98	1.72	1.52	1.40
788-8002AI133E-14.7T	SIT8002AI-13-33E-14.74560T	14.74560	1.98	1.72	1.52	1.40
788-8002AI133E-16.0T	SIT8002AI-13-33E-16.00000T	16.00000	1.98	1.72	1.52	1.40
788-8002AI133E-18.4T	SIT8002AI-13-33E-18.43200T	18.43200	1.98	1.72	1.52	1.40
788-8002AI133E-20.0T	SIT8002AI-13-33E-20.00000T	20.00000	1.98	1.72	1.52	1.40
788-8002AI133E-24.0T	SIT8002AI-13-33E-24.00000T	24.00000	1.98	1.72	1.52	1.40
788-8002AI133E-24.5T	SIT8002AI-13-33E-24.57600T	24.57600	1.98	1.72	1.52	1.40
788-8002AI133E-25.0T	SIT8002AI-13-33E-25.00000T	25.00000	1.98	1.72	1.52	1.40
788-8002AI133E-27.0T	SIT8002AI-13-33E-27.00000T	27.00000	1.98	1.72	1.52	1.40
788-8002AI133E-30.0T	SIT8002AI-13-33E-30.00000T	30.00000	1.98	1.72	1.52	1.40
788-8002AI133E-32.0T	SIT8002AI-13-33E-32.00000T	32.00000	1.98	1.72	1.52	1.40
788-8002AI133E-32.7T	SIT8002AI-13-33E-32.76800T	32.76800	1.98	1.72	1.52	1.40
788-8002AI133E-33.0T	SIT8002AI-13-33E-33.00000T	33.00000	1.98	1.72	1.52	1.40
788-8002AI133E-33.3T	SIT8002AI-13-33E-33.33300T	33.33300	1.98	1.72	1.52	1.40
788-8002AI133E-40.0T	SIT8002AI-13-33E-40.00000T	40.00000	1.98	1.72	1.52	1.40
788-8002AI133E-48.0T	SIT8002AI-13-33E-48.00000T	48.00000	1.98	1.72	1.52	1.40
788-8002AI133E-50.0T	SIT8002AI-13-33E-50.00000T	50.00000	1.98	1.72	1.52	1.40
788-8002AI133E-60.0T	SIT8002AI-13-33E-60.00000T	60.00000	1.98	1.72	1.52	1.40
788-8002AI133E-64.0T	SIT8002AI-13-33E-64.00000T	64.00000	1.98	1.72	1.52	1.40
788-8002AI133E-66.0T	SIT8002AI-13-33E-66.00000T	66.00000	1.98	1.72	1.52	1.40
788-8002AI133E-66.6T	SIT8002AI-13-33E-66.66667T	66.66667	1.98	1.72	1.52	1.40
788-8002AI133E-75.0T	SIT8002AI-13-33E-75.00000T	75.00000	1.98	1.72	1.52	1.40
788-8002AI133E-80.0T	SIT8002AI-13-33E-80.00000T	80.00000	1.98	1.72	1.52	1.40
788-8002AI133E-100T	SIT8002AI-13-33E-100.00000T	100.0000	2.16	1.87	1.66	1.53
788-8002AI133E-106T	SIT8002AI-13-33E-106.25000T	106.2500	2.16	1.87	1.66	1.53
788-8002AI133E-125T	SIT8002AI-13-33E-125.00000T	125.0000	2.16	1.87	1.66	1.53

- Other frequencies and voltages available (1.8V and 2.5V).

SIT8002 SERIES: 3.3 VOLT, 3.2MM X 2.5MM PACKAGE

Features:
- One Time Programmable
- 1 MHz to 125 MHz Clock
- Ultra short leadtime
- Greater immunity from interference and ultra-reliable start up

PINOUT:
Pad 1: Standby, Output Enable
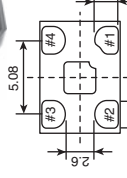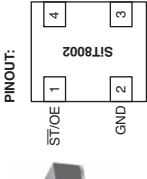Pad 2: Connect to Ground
Pad 3: Clock Output
Pad 4: Connect +3.3V

For quantities of 250 and up, call for quote.

MOUSER STOCK NO.	SiTime Part No.	Freq. (MHz)	Price Each 1	10	50	100
788-8002AI233E-2.04T	SIT8002AI-23-33E-2.04800T	2.04800	1.83	1.59	1.40	1.30
788-8002AI233E-4.0T	SIT8002AI-23-33E-4.00000T	4.00000	1.83	1.59	1.40	1.30
788-8002AI233E-8.0T	SIT8002AI-23-33E-8.00000T	8.00000	1.83	1.59	1.40	1.30
788-8002AI233E-10.0T	SIT8002AI-23-33E-10.00000T	10.00000	1.83	1.59	1.40	1.30
788-8002AI233E-12.0T	SIT8002AI-23-33E-12.00000T	12.00000	1.83	1.59	1.40	1.30
788-8002AI233E-13.5T	SIT8002AI-23-33E-13.56000T	13.56000	1.83	1.59	1.40	1.30
788-8002AI233E-14.3T	SIT8002AI-23-33E-14.31818T	14.31818	1.83	1.59	1.40	1.30
788-8002AI233E-16.0T	SIT8002AI-23-33E-16.00000T	16.00000	1.83	1.59	1.40	1.30
788-8002AI233E-20.0T	SIT8002AI-23-33E-20.00000T	20.00000	1.83	1.59	1.40	1.30
788-8002AI233E-24.0T	SIT8002AI-23-33E-24.00000T	24.00000	1.83	1.59	1.40	1.30
788-8002AI233E-24.5T	SIT8002AI-23-33E-24.57600T	24.57600	1.83	1.59	1.40	1.30
788-8002AI233E-25.0T	SIT8002AI-23-33E-25.00000T	25.00000	1.83	1.59	1.40	1.30
788-8002AI233E-27.0T	SIT8002AI-23-33E-27.00000T	27.00000	1.83	1.59	1.40	1.30
788-8002AI233E-33.3T	SIT8002AI-23-33E-33.33300T	33.33300	1.83	1.59	1.40	1.30
788-8002AI233E-40.0T	SIT8002AI-23-33E-40.00000T	40.00000	1.83	1.59	1.40	1.30
788-8002AI233E-48.0T	SIT8002AI-23-33E-48.00000T	48.00000	1.83	1.59	1.40	1.30
788-8002AI233E-50.0T	SIT8002AI-23-33E-50.00000T	50.00000	1.83	1.59	1.40	1.30
788-8002AI233E-75.0T	SIT8002AI-23-33E-75.00000T	75.00000	2.01	1.74	1.54	1.42
788-8002AI233E-80.0T	SIT8002AI-23-33E-80.00000T	80.00000	2.01	1.74	1.54	1.42
788-8002AI233E-100T	SIT8002AI-23-33E-100.00000T	100.0000	2.01	1.74	1.54	1.42
788-8002AI233E-106T	SIT8002AI-23-33E-106.25000T	106.2500	2.01	1.74	1.54	1.42
788-8002AI233E-125T	SIT8002AI-23-33E-125.00000T	125.0000	2.01	1.74	1.54	1.42

- Other frequencies and voltages available (1.8V and 2.5V).

SIT8002 SERIES: 3.3 VOLT, 7.0MM X 5.0MM PACKAGE

Features:
- One Time Programmable
- 1 MHz to 125 MHz Clock
- Ultra short leadtime
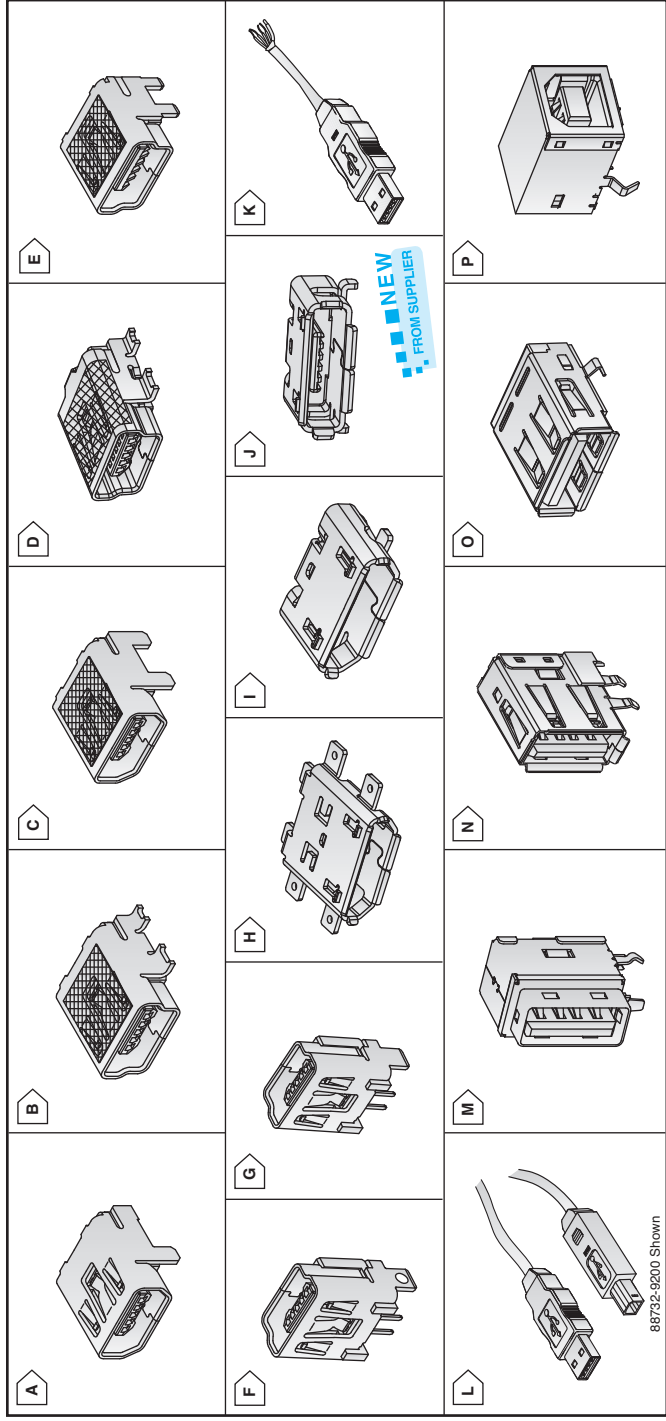- Greater immunity from interference and ultra-reliable start up

PINOUT:
Pad 1: Standby, Output Enable
Pad 2: Connect to Ground
Pad 3: Clock Output
Pad 4: Connect +3.3V
The center pad is No Connect (NC)

For quantities of 250 and up, call for quote.

MOUSER STOCK NO.	SiTime Part No.	Freq. (MHz)	Price Each 1	10	50	100
788-8002AI433E-1.84T	SIT8002AI-43-33E-1.84320T	1.84320	1.62	1.40	1.24	1.15
788-8002AI433E-2.04T	SIT8002AI-43-33E-2.04800T	2.04800	1.62	1.40	1.24	1.15
788-8002AI433E-3.68T	SIT8002AI-43-33E-3.68640T	3.68640	1.62	1.40	1.24	1.15
788-8002AI433E-4.0T	SIT8002AI-43-33E-4.00000T	4.00000	1.62	1.40	1.24	1.15
788-8002AI433E-5.0T	SIT8002AI-43-33E-5.00000T	5.00000	1.62	1.40	1.24	1.15
788-8002AI433E-6.0T	SIT8002AI-43-33E-6.00000T	6.00000	1.62	1.40	1.24	1.15
788-8002AI433E-6.14T	SIT8002AI-43-33E-6.14000T	6.14000	1.62	1.40	1.24	1.15
788-8002AI433E-8.0T	SIT8002AI-43-33E-8.00000T	8.00000	1.62	1.40	1.24	1.15
788-8002AI433E-10.0T	SIT8002AI-43-33E-10.00000T	10.00000	1.62	1.40	1.24	1.15
788-8002AI433E-12.0T	SIT8002AI-43-33E-12.00000T	12.00000	1.62	1.40	1.24	1.15
788-8002AI433E-13.0T	SIT8002AI-43-33E-13.00000T	13.00000	1.62	1.40	1.24	1.15
788-8002AI433E-13.5T	SIT8002AI-43-33E-13.56000T	13.56000	1.62	1.40	1.24	1.15
788-8002AI433E-14.3T	SIT8002AI-43-33E-14.31818T	14.31818	1.62	1.40	1.24	1.15
788-8002AI433E-14.7T	SIT8002AI-43-33E-14.74560T	14.74560	1.62	1.40	1.24	1.15
788-8002AI433E-16.0T	SIT8002AI-43-33E-16.00000T	16.00000	1.62	1.40	1.24	1.15
788-8002AI433E-18.4T	SIT8002AI-43-33E-18.43200T	18.43200	1.62	1.40	1.24	1.15
788-8002AI433E-20.0T	SIT8002AI-43-33E-20.00000T	20.00000	1.62	1.40	1.24	1.15
788-8002AI433E-24.0T	SIT8002AI-43-33E-24.00000T	24.00000	1.62	1.40	1.24	1.15
788-8002AI433E-24.5T	SIT8002AI-43-33E-24.57600T	24.57600	1.62	1.40	1.24	1.15
788-8002AI433E-25.0T	SIT8002AI-43-33E-25.00000T	25.00000	1.62	1.40	1.24	1.15
788-8002AI433E-27.0T	SIT8002AI-43-33E-27.00000T	27.00000	1.62	1.40	1.24	1.15
788-8002AI433E-30.0T	SIT8002AI-43-33E-30.00000T	30.00000	1.62	1.40	1.24	1.15
788-8002AI433E-32.7T	SIT8002AI-43-33E-32.76800T	32.76800	1.62	1.40	1.24	1.15
788-8002AI433E-33.0T	SIT8002AI-43-33E-33.00000T	33.00000	1.62	1.40	1.24	1.15
788-8002AI433E-33.3T	SIT8002AI-43-33E-33.33300T	33.33300	1.62	1.40	1.24	1.15
788-8002AI433E-40.0T	SIT8002AI-43-33E-40.00000T	40.00000	1.62	1.40	1.24	1.15
788-8002AI433E-48.0T	SIT8002AI-43-33E-48.00000T	48.00000	1.62	1.40	1.24	1.15
788-8002AI433E-50.0T	SIT8002AI-43-33E-50.00000T	50.00000	1.62	1.40	1.24	1.15
788-8002AI433E-60.0T	SIT8002AI-43-33E-60.00000T	60.00000	1.62	1.40	1.24	1.15
788-8002AI433E-66.0T	SIT8002AI-43-33E-66.00000T	66.00000	1.62	1.40	1.24	1.15
788-8002AI433E-66.6T	SIT8002AI-43-33E-66.66667T	66.66667	1.62	1.40	1.24	1.15
788-8002AI433E-80.0T	SIT8002AI-43-33E-80.00000T	80.00000	1.80	1.56	1.38	1.28
788-8002AI433E-100T	SIT8002AI-43-33E-100.00000T	100.0000	1.80	1.56	1.38	1.28
788-8002AI433E-106T	SIT8002AI-43-33E-106.25000T	106.2500	1.80	1.56	1.38	1.28
788-8002AI433E-125T	SIT8002AI-43-33E-125.00000T	125.0000	1.80	1.56	1.38	1.28

- Other frequencies and voltages available (1.8V and 2.5V).

SIT8002 SERIES: 1.8 VOLT, 2.5MM X 2.0MM PACKAGE

Features:
- One Time Programmable
- 1 MHz to 125 MHz Clock
- Ultra short leadtime
- Greater immunity from interference and ultra-reliable start up

PINOUT:
Pad 1: Standby, Output Enable
Pad 2: Connect to Ground
Pad 3: Clock Output
Pad 4: Connect +1.8V

For quantities of 250 and up, call for quote.

MOUSER STOCK NO.	SiTime Part No.	Freq. (MHz)	Price Each 1	10	50	100
788-8002AI131E-2.04T	SIT8002AI-13-18E-2.04800T	2.04800	2.07	1.79	1.59	1.47
788-8002AI131E-4.0T	SIT8002AI-13-18E-4.00000T	4.00000	2.07	1.79	1.59	1.47
788-8002AI131E-8.0T	SIT8002AI-13-18E-8.00000T	8.00000	2.07	1.79	1.59	1.47
788-8002AI131E-12.0T	SIT8002AI-13-18E-12.00000T	12.00000	2.07	1.79	1.59	1.47
788-8002AI131E-14.3T	SIT8002AI-13-18E-14.31818T	14.31818	2.07	1.79	1.59	1.47
788-8002AI131E-16.0T	SIT8002AI-13-18E-16.00000T	16.00000	2.07	1.79	1.59	1.47
788-8002AI131E-20.0T	SIT8002AI-13-18E-20.00000T	20.00000	2.07	1.79	1.59	1.47
788-8002AI131E-24.0T	SIT8002AI-13-18E-24.00000T	24.00000	2.07	1.79	1.59	1.47
788-8002AI131E-24.5T	SIT8002AI-13-18E-24.57600T	24.57600	2.07	1.79	1.59	1.47
788-8002AI131E-25.0T	SIT8002AI-13-18E-25.00000T	25.00000	2.07	1.79	1.59	1.47
788-8002AI131E-27.0T	SIT8002AI-13-18E-27.00000T	27.00000	2.07	1.79	1.59	1.47
788-8002AI131E-33.3T	SIT8002AI-13-18E-33.33300T	33.33300	2.07	1.79	1.59	1.47
788-8002AI131E-40.0T	SIT8002AI-13-18E-40.00000T	40.00000	2.07	1.79	1.59	1.47
788-8002AI131E-48.0T	SIT8002AI-13-18E-48.00000T	48.00000	2.07	1.79	1.59	1.47
788-8002AI131E-50.0T	SIT8002AI-13-18E-50.00000T	50.00000	2.07	1.79	1.59	1.47
788-8002AI131E-80.0T	SIT8002AI-13-18E-80.00000T	80.00000	2.16	1.87	1.66	1.53
788-8002AI131E-100T	SIT8002AI-13-18E-100.00000T	100.0000	2.16	1.87	1.66	1.53
788-8002AI131E-106T	SIT8002AI-13-18E-106.25000T	106.2500	2.16	1.87	1.66	1.53
788-8002AI131E-125T	SIT8002AI-13-18E-125.00000T	125.0000	2.16	1.87	1.66	1.53

- Other frequencies and voltages available (1.8V and 2.5V).

Appendix E-4: USB Connector Specifications

MOLEX USB Connectors and Cable Assemblies

88732-9200 Shown

USB ON-THE-GO (OTG) CONNECTORS AND CABLE ASSEMBLIES

Molex's new USB On-The-Go (OTG) connectors and cable assemblies help enable mobile devices such as digital cameras, PDAs, audio players, printers and cellular phones to exchange data directly without the need for a host PC. All Molex OTG connectors are compatible with the On-The-Go supplement (Rev. 1.0) to the USB 2.0 specification. Along with enabling mobile products to communicate with one another, OTG also enables mobile products to become "dual-role" devices, serving as either host or peripheral.

Features:
- Compact size for space savings
- Operates at speeds up to 480Mbps
- Fully shielded for EMI/RFI protection
- 5000 cycles for rugged durability

Specifications:
- Current: 1.0A
- Voltage Rating: 30V / Micro: 100VAC (RMS)
- Operating Temperature: 0°C to +50°C

RoHS Compliant

Mini-AB Receptacles

MOUSER STOCK NO.	Molex Part No.	Fig.	No. of Circuits	Description	Shell Tabs	Solder Tail	1	10	25	50
538-56579-0519	56579-0519	A	5	Right-Angle	Through-Hole	Through-Hole	3.58	2.81	2.50	2.26
538-56579-0576	56579-0576	B	5	Right-Angle	SMT	SMT	3.12	2.43	2.09	1.92

Mini-B Receptacles

MOUSER STOCK NO.	Molex Part No.	Fig.	No. of Circuits	Description	Shell Tabs	Solder Tail	1	10	25	50
538-54819-0519	54819-0519	A	5	Right-Angle	Through-Hole	Through-Hole	2.05	1.71	1.42	1.24
538-54819-0589	54819-0589	C	5	Right-Angle	Through-Hole	SMT	1.54	1.41	1.32	1.20
538-54819-0572	54819-0572	B	5	Right-Angle	SMT	SMT	1.56	1.30	1.17	1.05
538-67503-1020	67503-1020	D	5	Right-Angle (Back Cover)	SMT	SMT	1.12	.94	.82	.75
538-51387-0578	51387-0578	E	5	Right-Angle	Through-Hole	SMT	1.74	1.52	1.40	1.26
538-500075-0517	500075-0517	F	5	Vertical	Through-Hole	Through-Hole	6.03	4.83	4.40	3.82
538-500075-1517	500075-1517	G	5	Vertical	Through-Hole	Through-Hole	5.77	4.62	4.04	3.66

Micro-USB Receptacles

MOUSER STOCK NO.	Molex Part No.	Fig.	No. of Circuits	Description	Shell Tabs	Solder Tail	1	10	25	50
Type B										
538-47491-0001	47491-0001	H	5	Right-Angle	Middle Mount	SMT	1.22	1.12	1.00	.92
538-47346-0001	47346-0001	I	5	Right-Angle	Bottom Mount	SMT	1.07	.99	.92	.84
Type AB										
538-47589-0001	47589-0001	J	5	Right-Angle	Bottom Mount	SMT	.79	.73	.66	.59
538-47590-0001	47590-0001	J	5	Right-Angle	Top Mount	SMT	.79	.73	.66	.59

Cable Assemblies

MOUSER STOCK NO.	Molex Part No.	Fig.	No. of Circuits	Type	Description	1	10	25	50
538-88738-8200	88738-8200	K	4	USB Type A-to-Pigtail	2.08m Length, Sub-Channel	3.17	2.51	2.04	1.85
538-88728-3400	88728-3400	K	4	USB Type A-to-Pigtail	2.08m Length, Full-Rated	4.14	3.32	2.90	2.62
538-68784-0001	68784-0001	L	–	USB Type A-to-Micro B	1 Meter Length, 28 AWG	4.42	3.97	3.58	3.22
538-68784-0002	68784-0002	L	–	USB Type A-to-Micro B	1.5 Meter Length, 28 AWG	4.99	4.51	4.05	3.64
538-68784-0003	68784-0003	L	–	USB Type A-to-Micro B	2 Meter Length, 28 AWG	5.57	4.98	4.51	4.06
538-88732-9200	88732-9200	L	4	USB Type A-to-B	2.09m Length, Full-Rated	4.11	3.78	3.26	3.00

USB TYPE "A" AND "B" RECEPTACLES

The basic characteristics of all 'A' and 'B' receptacles are: having a metal shell to provide for excellent EMI/RFI protection, PCB hold-downs meet specifications of an electrical rating of 1.0 Amp per contact and are designed to endure at least 1,500 mating cycles.

MOUSER STOCK NO.	Molex Part No.	Fig.	No. of Circuits	Type	Description	1	10	25	50
538-89485-8000	89485-8000	M	4	Type A	Single Upright, Right-Angle, Black, Gold Flash Plating	1.30	1.22	1.12	1.00
538-89485-8001	89485-8001	M	4	Type A	Single Upright, Right-Angle, Black, Black 30μ"Gold Plating	2.07	1.68	1.44	1.23
538-67329-8000	67329-8000	N	4	Type A	Single Upright, Right Angle, Gold Flash Plating	1.85	1.63	1.16	1.06
538-67643-2911	67643-2911	O	4	Type A	Right Angle, High Temperature Black nylon, 30μ" Gold Plating	1.99	1.84	1.69	1.58
538-67643-3911	67643-3911	O	4	Type A	Right Angle, High Temperature White nylon, 30μ" Gold Plating	1.99	1.83	1.68	1.53
538-67068-8000	67068-8000	P	4	Type B	Right-Angle, w/ Kinked Solder Tabs, Black, Gold Flash Plating	1.78	1.47	1.30	1.18
538-67068-8001	67068-8001	P	4	Type B	Right-Angle, w/ Kinked Solder Tabs, Black, 30μ" Gold Plating	1.82	1.57	1.40	1.26
538-67068-9000	67068-9000	P	4	Type B	Right-Angle, w/ Kinked Solder Tabs, White, Gold Flash Plating	1.45	1.31	1.17	1.05
538-67068-9001	67068-9001	P	4	Type B	Right-Angle, w/ Kinked Solder Tabs, White, 30μ" Gold Plating	1.72	1.48	1.40	1.32

NEW FROM SUPPLIER

Mouser Catalog Download

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Appendix E-5: Schottky Diode Specifications

New Product

MSS1P3 & MSS1P4

Vishay General Semiconductor

Surface Mount Schottky Barrier Rectifiers

eSMP™ Series

MicroSMP

Top View   Bottom View
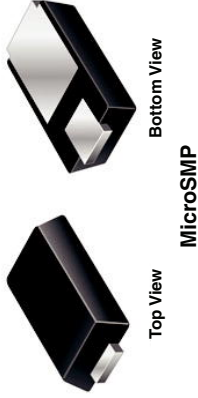
FEATURES

- Very low profile - typical height of 0.68 mm
- Ideal for automated placement
- Low forward voltage drop, low power losses
- High efficiency
- Meets MSL level 1, per J-STD-020, LF maximum peak of 260 °C
- Component in accordance to RoHS 2002/95/EC and WEEE 2002/96/EC
- **Halogen-free**

MECHANICAL DATA

Case: MicroSMP

Molding compound meets UL 94V-0 flammability rating.

Base P/N-E3 - RoHS compliant, commercial grade

Base P/N-M3 - halogen-free and RoHS compliant, commercial grade

Terminals: Matte tin plated leads, solderable per J-STD-002 and JESD22-B102

E3 and M3 suffix meets JESD 201 class 1A whisker test

Polarity: Color band denotes the cathode end

PRIMARY CHARACTERISTICS

$I_{F(AV)}$	1.0 A
V_{RRM}	30 V, 40 V
I_{FSM}	25 A
V_F at I_F = 1.0 A	0.41 V
T_J max.	150 °C

TYPICAL APPLICATIONS

For use in low voltage high frequency inverters, freewheeling, dc-to-dc converters, and polarity protection applications.

MAXIMUM RATINGS (T_A = 25 °C unless otherwise noted)

PARAMETER	SYMBOL	MSS1P3	MSS1P4	UNIT
Device marking code		13	14	
Maximum repetitive peak reverse voltage	V_{RRM}	30	40	V
Maximum average forward rectified current (Fig. 1)	$I_{F(AV)}$	1.0		A
Peak forward surge current 8.3 ms single half sine-wave superimposed on rated load	I_{FSM}	25		A
Operating junction and storage temperature range	T_J, T_{STG}	- 55 to + 150		°C

MSS1P3 & MSS1P4

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New Product

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ELECTRICAL CHARACTERISTICS (T$_A$ = 25 °C unless otherwise noted)

PARAMETER	TEST CONDITIONS		SYMBOL	TYP.	MAX.	UNIT
Maximum instantaneous forward voltage [1]	I$_F$ = 0.5 A I$_F$ = 1.0 A	T$_J$ = 25 °C	V$_F$	0.41 0.48	- 0.55	V
	I$_F$ = 0.5 A I$_F$ = 1.0 A	T$_J$ = 125 °C		0.32 0.41	- 0.46	
Maximum reverse current [2]	rated V$_R$	T$_J$ = 25 °C T$_J$ = 125 °C	I$_R$	8.5 4.5	200 15	µA mA
Typical junction capacitance	4.0 V, 1 MHz		C$_J$	50	-	pF

Notes:
(1) Pulse test: 300 µs pulse width, 1 % duty cycle
(2) Pulse test: Pulse width ≤ 40 ms

THERMAL CHARACTERISTICS (T$_A$ = 25 °C unless otherwise noted)

PARAMETER	SYMBOL	MSS1P3	MSS1P4	UNIT
Typical thermal resistance [1]	R$_{\theta JA}$ R$_{\theta JL}$ R$_{\theta JC}$	125 30 40		°C/W

Note:
(1) Thermal resistance from junction to ambient and junction to lead mounted on P.C.B. with 6.0 x 6.0 mm copper pad areas
R$_{\theta JL}$ is measured at the terminal of cathode band. R$_{\theta JC}$ is measured at the top center of the body

ORDERING INFORMATION (Example)

PREFERRED P/N	UNIT WEIGHT (g)	PREFERRED PACKAGE CODE	BASE QUANTITY	DELIVERY MODE
MSS1P4-E3/89A	0.006	89A	4500	7" diameter plastic tape and reel
MSS1P4-M3/89A	0.006	89A	4500	7" diameter plastic tape and reel

RATINGS AND CHARACTERISTICS CURVES
(T$_A$ = 25 °C unless otherwise noted)
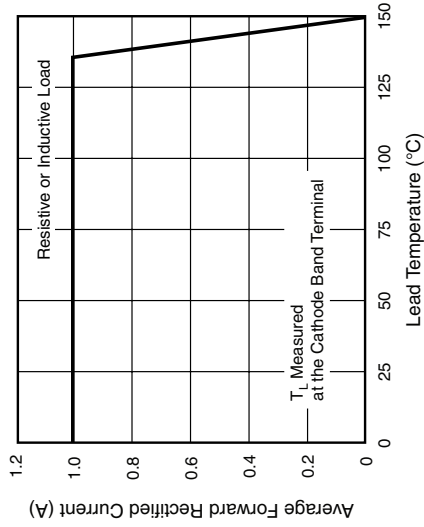
Figure 1. Maximum Forward Current Derating Curve
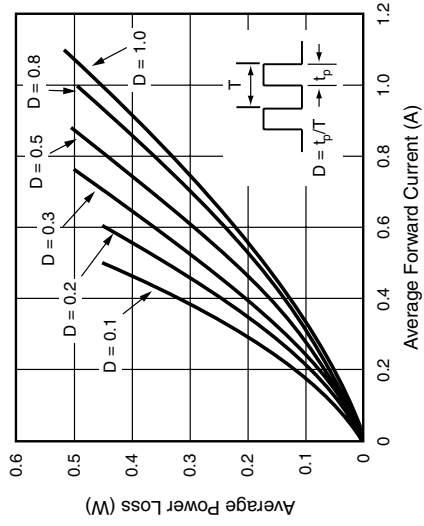
Figure 2. Forward Power Loss Characteristics

MSS1P3 & MSS1P4

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Figure 5. Typical Junction Capacitance

Figure 6. Typical Transient Thermal Impedance
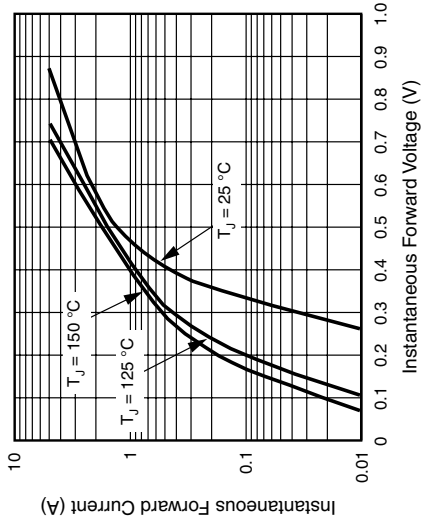
Figure 3. Typical Instantaneous Forward Characteristics
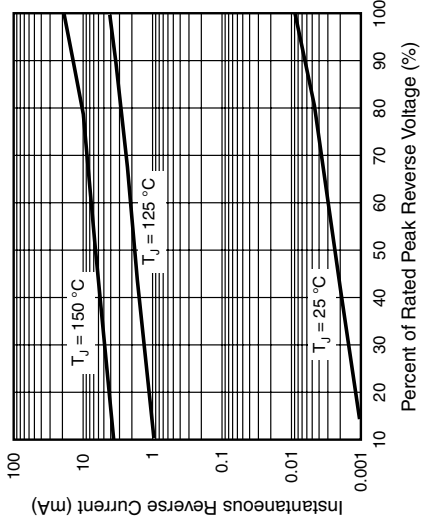
Figure 4. Typical Reverse Characteristics

PACKAGE OUTLINE DIMENSIONS in inches (millimeters)

MicroSMP

Mounting Pad Layout

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Appendix E-6: Linear Regulator Specifications

High-Performance, Size Saving 150-mA CMOS LDO Regulator

FEATURES

- Low 135-mV Dropout at 150-mA Load
- Guaranteed 150-mA Output Current
- 300-mA Peak Output Current Capability
- Uses Low ESR Ceramic Output Capacitor
- Fast Load And Line Transient Response
- Low Output Noise
- 1-μA Maximum Shutdown Current
- Built-in Short Circuit And Thermal Protection
- Fixed 1.8-V, 2.5-V, 2.8-V, 2.85-V, 3.0-V, 3.3-V, 5.0-V or Adjustable Output Voltage Options (Version B)

- Thin SOT-23 5-Pin Package

Pb-free
Available

APPLICATIONS

- Battery Powered Portable Systems
- Cellular Phones
- PDAs, Palmtops
- Pagers
- Post Regulators for Multi-Output Converters
- Notebook Computers

DESCRIPTION

The Si9183 is a high performance yet size saving 150-mA CMOS LDO (low dropout) voltage regulator. Its ultra low ground current and dropout voltage prolong battery life in portable electronics. The device provides LINE/LOAD transient response and ripple rejection superior to that of Bipolar or BiCMOS LDO regulators. It is designed to maintain regulation while delivering 300-mA peak current. The Si9183 drives lower cost ceramic, as well as tantalum, output capacitors. Stability is guaranteed from maximum load current down to 0-mA load. An external noise bypass capacitor connected to the device's C$_{BP}$ pin will reduce the LDO's

self-noise for low noise applications. The Si9183 includes a shutdown feature that allows users to completely disable the device and save power when no output is required.

The Si9183, in Thin SOT23-5 packaging, is available in two versions (Version A or B). Version A offers low noise performance, while Version B features adjustable output voltage.

The Si9183 is available in both standard and lead (Pb)-free packages.

TYPICAL APPLICATIONS CIRCUITS

Si9183-A

Thin SOT-23, 5-Lead
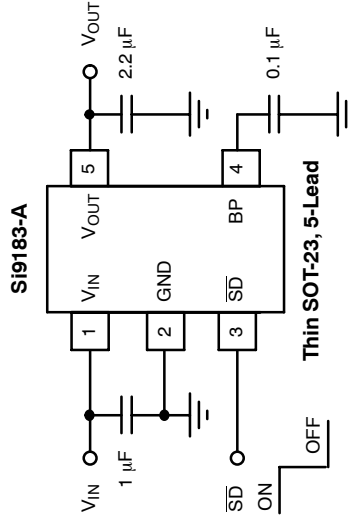
FIGURE 1. Version A with Low Output Noise

Si9183-B

Thin SOT-23, 5-Lead
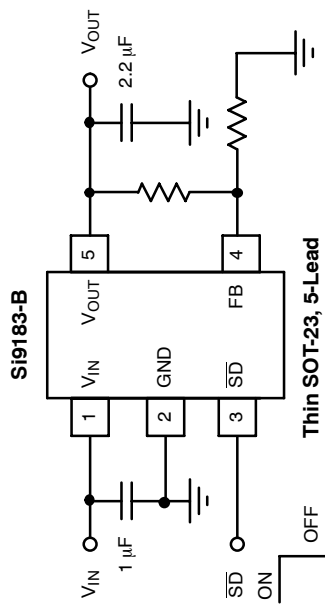
FIGURE 2. Version B with Adjustable Output

Si9183

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ABSOLUTE MAXIMUM RATINGS

Input Voltage, V_{IN} .... 6.5 V
$\overline{SD}$ Input Voltage, $V_{\overline{SD}}$ .... −0.3 V to V_{IN}
Output Current, I_{OUT} .... Short Circuit Protected
Output Voltage, V_{OUT} .... −0.3 V to $V_{O(nom)}$ + 0.3 V
Maximum Junction Temperature, $T_{J(max)}$ .... 150°C
Storage Temperature, T_{STG} .... −65°C to 125°C
ESD (Human Body Model) .... 2 kV

Power Dissipation (Package)a, b
5-Pin SOT-23 .... 555 mW
Thermal Impedance (Θ_{JA})
5-Pin SOT-23 .... 180 °C/W

Notes
a. Device mounted with all leads soldered or welded to multi-layer (1S2P) JEDEC board, horizontal orientation.
b. Derate 5.5 mW/°C above T_A = 25°C.

Stresses beyond those listed under "Absolute Maximum Ratings" may cause permanent damage to the device. These are stress ratings only, and functional operation of the device at these or any other conditions beyond those indicated in the operational sections of the specifications is not implied. Exposure to absolute maximum rating conditions for extended periods may affect device reliability.

RECOMMENDED OPERATING RANGE

Input Voltage, V_{IN} .... 2 V to 6 V
Output Voltage, V_{OUT} (Adjustable Version) .... 1.5 V to 5 V
$\overline{SD}$ Input Voltage, $V_{\overline{SD}}$ .... 0 V to V_{IN}

Operating Ambient Temperature, T_A .... −40°C to 85°C
Operating Junction Temperature, T_J .... −40°C to 125°C

C_{IN} = 1 µF, C_{OUT} = 2.2 µF (ceramic, X5R or X7R type) , C_{BP} = 0.1 µF (ceramic)
C_{OUT}Range = 1 µF to 10 µF (±20% tolerance, ±20% over temperature; ESR = 0.4 to 4 Ω at dc to 100 kHz, 0 t0 0.4 Ω > 100 kHz))

SPECIFICATIONS (T_A = 25°C)

Parameter	Symbol	Test Conditions Unless Otherwise Specified	Tempa	Limits −40 to 85°C Minb	Typc	Maxb	Unit
Input Voltage Range	V_{IN}		Full	2		6	V
Output Voltage Range	V_{OUT}	Adjustable Version	Full	1.5		5	
Output Voltage Accuracy (Fixed Versions)		1 mA ≤ I_{OUT} ≤ 150 mA	Room	−1.5		1.5	% $V_{O(nom)}$
			Full	−2.5		2.5	
Feedback Voltage (ADJ version)	V_{FB}		Room	1.188	1.215	1.240	V
			Full	1.176		1.252	
Line Regulation (Except 5-V Version)	$\dfrac{\Delta V_{OUT} \times 100}{V_{IN} \times V_{OUT(nom)}}$	From V_{IN} = $V_{OUT(nom)}$ + 1 V to $V_{OUT(nom)}$ + 2 V	Full	−0.18		0.18	%/V
Line Regulation (5-V Version)		From V_{IN} = 5.5 V to 6 V	Full	−0.18		0.18	
Line Regulation (ADJ Version)		V_{OUT} = 1.5 V, From V_{IN} = 2.5 V to 3.5 V	Full	−0.18		0.18	
		V_{OUT} = 5 V, From V_{IN} = 5.5 V to 6 V	Full	−0.18		0.18	
Dropout Voltaged @V_{OUT} ≥ 2.5 V	V_{IN} − V_{OUT}	I_{OUT} = 10 mA	Room		1	20	mV
		I_{OUT} = 150 mA	Room		135	170	
			Full		180	220	
Dropout Voltaged (@V_{OUT} < 2.5 V, V_{IN} ≥ 2 V)	V_{IN} − V_{OUT}	I_{OUT} = 150 mA	Room		235	320	
			Full			380	
Ground Pin Current	I_{GND}	I_{OUT} = 0 mA	Room		150		µA
		I_{OUT} = 150 mA	Room		500		
			Full			900	
Shutdown Supply Current	$I_{IN(off)}$	$V_{\overline{SD}}$ = 0 V	Full		0.1	1	µA
FB Pin Current	I_{FB}	V_{FB} = 1.2 V	Room		2	100	nA
Peak Output Current	$I_{O(peak)}$	V_{OUT} ≥ 0.95 × $V_{OUT(nom)}$, t_{pw} = 2 ms	Room	250	300		mA
Output Noise Voltage	e_N	BW = 50 Hz to 100 kHz I_{OUT} = 150 mA, w/o C_{BP}	Room		300		µV (rms)
		C_{BP} = 0.1 µF	Room		100		

SPECIFICATIONS (T_A = 25°C)

Parameter	Symbol	Test Conditions Unless Otherwise Specified $V_{IN} = V_{OUT(nom)} + 1\ V$, $I_{OUT} = 1\ mA$ $C_{IN} = 1\ \mu F$, $C_{OUT} = 2.2\ \mu F$, $\overline{V_{SD}} = 1.5\ V$		Temp[a]	Limits −40 to 85°C			Unit
					Min[b]	Typ[c]	Max[b]	
Ripple Rejection	$\Delta V_{OUT}/\Delta V_{IN}$	$I_{OUT} = 150\ mA$	f = 1 kHz	Room		60		dB
			f = 10 kHz	Room		40		
			f = 100 kHz	Room		30		
Dynamic Line Regulation	$\Delta V_{O(line)}$	$V_{IN} : V_{OUT(nom)} + 1\ V$ to $V_{OUT(nom)} + 2\ V$ $t_R/t_F = 5\ \mu s$, $I_{OUT} = 150\ mA$		Room		10		mV
Dynamic Load Regulation	$\Delta V_{O(load)}$	$I_{OUT} : 1\ mA$ to $150\ mA$, $t_R/t_F = 2\ \mu s$		Room		30		
V_{OUT} Turn-On-Time	t_{ON}	$V_{IN} = 4.3\ V$ $V_{OUT} = 3.3\ V$	w/o C_{BP} Cap	Room		5		µs
			$C_{BP} = 0.1\ \mu F$	Room		1000		
Thermal Shutdown								
Thermal Shutdown Junction Temp	$t_{J(s/d)}$			Room		165		°C
Thermal Hysteresis	t_{HYST}			Room		20		
Short Circuit Current	I_{SC}	$V_{OUT} = 0\ V$		Room		400		mA
Shutdown Input								
$\overline{SD}$ Input Voltage	V_{IH}	High = Regulator ON (Rising)		Full	1.2		V_{IN}	V
	V_{IL}	Low = Regulator OFF (Falling)		Full			0.4	
$\overline{SD}$ Input Current[e]	I_{IL}	$\overline{V_{SD}} = 0\ V$, Regulator OFF		Room		0.01		µA
	I_{IH}	$\overline{V_{SD}} = 6\ V$, Regulator ON		Room		1.0		
Shutdown Hysteresis	V_{HYST}			Full		100		mV

Notes
a. Room = 25°C, Full = −40 to 85°C.
b. The algebraic convention whereby the most negative value is a minimum and the most positive a maximum.
c. Typical values are for DESIGN AID ONLY, not guaranteed nor subject to production testing. Typical values for dropout voltage at $V_{OUT} \geq 2\ V$ are measured at $V_{OUT} = 2.5\ V$, while typical values for dropout voltage at $V_{OUT} < 2\ V$ are measured at $V_{OUT} = 1.8\ V$.
d. Dropout voltage is defined as the input to output differential voltage at which the output voltage drops 2% below the output voltage measured with a 1-V differential, provided that V_{IN} does not drop below 2.0 V.
e. The device's shutdown pin includes a typical 6-MΩ internal pull-down resistor connected to ground.
f. V_{OUT} is defined as the output voltage of the DUT at 1 mA.

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TIMING WAVEFORMS
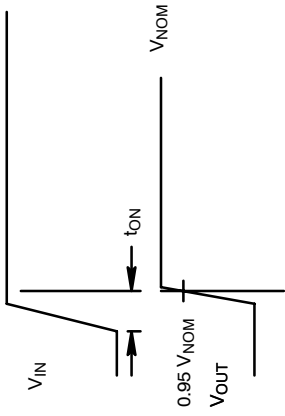
FIGURE 3. Timing Diagram for Power-Up

PIN CONFIGURATION

Thin SOT-23, 5-Pin
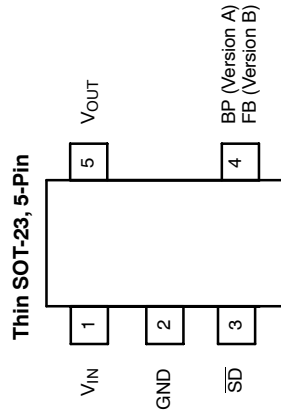
PIN DESCRIPTION

Pin Number	Name	Function
1	V_{IN}	Input supply pin. Bypass this pin with a 1-μF ceramic or tantalum capacitor to ground.
2	GND	Ground pin. Local ground for C_{BP} and C_{OUT}.
3	$\overline{SD}$	By applying less than 0.4 V to this pin, the device will be turned off. Connect this pin to V_{IN} if unused.
4 (Version A)	BP	Noise bypass pin. For low noise applications, a 0.1-μF or larger ceramic capacitor should be connected from this pin to ground.
4 (Version B)	FB	Connect to divided output voltage to adjust the regulation point.
5	V_{OUT}	Output voltage. Connect C_{OUT} between this pin and ground.

ORDERING INFORMATION

Part Number	Lead (Pb)-Free Part Number	Marking	Voltage	Temperature Range	Package
Si9183DT-18-T1	Si9183DT-18-T1—E3	A2LL	1.8 V		
Si9183DT-25-T1	Si9183DT-25-T1—E3	A4LL	2.5 V		
Si9183DT-28-T1	Si9183DT-28-T1—E3	A5LL	2.8 V		
Si9183DT-285-T1	Si9183DT-285-T1—E3	B3LL	2.85 V		Thin SOT23-5
Si9183DT-30-T1	Si9183DT-30-T1—E3	A6LL	3.0 V	–40 to 85°C	
Si9183DT-33-T1	Si9183DT-33-T1—E3	A7LL	3.3 V		
Si9183DT-50-T1	Si9183DT-50-T1—E3	A8LL	5.0 V		
Si9183DT-AD-T1	Si9183DT-AD-T1—E3	A9LL	Adjustable		

NOTE: LL = Lot Code

Eval Kit	Temperature Range	Board Type
Si9183DB	–40 to 85°C	Surface Mount

TYPICAL CHARACTERISTICS (INTERNALLY REGULATED, 25°C UNLESS NOTED)

V_IN – V_OUT Transfer Characteristic

Dropout Voltage vs. V_OUT

Dropout Voltage vs. Load Current

Dropout Voltage vs. Temperature

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TYPICAL CHARACTERISTICS (INTERNALLY REGULATED, 25°C UNLESS NOTED)

Normalized Output Voltage vs. Load Current

Normalized V_{OUT} vs. Temperature

No Load GND Pin Current vs. Input Voltage

GND Current vs. Load Current

Shutdown Threshold vs. Temperature

Shutdown Pin Input Current vs. Temperature

TYPICAL CHARACTERISTICS (INTERNALLY REGULATED, 25°C UNLESS NOTED)

Power Supply Rejection

$C_{IN} = 1 \ \mu F$
$C_{OUT} = 2.2 \ \mu F$
$I_{LOAD} = 150 \ mA$

Gain (dB) vs. Frequency (Hz)

I_{SC} vs. Temperature

$V_{OUT} = 2.5 \ V$

I_{SC} (A) vs. Ambient Temperature

TYPICAL WAVEFORMS

Turn-Off Delay

$\overline{SD}$ 2 V/div

V_{OUT} 2 V/div

5.00 μs/div

$V_{IN} = 4.2 \ V$
$V_{OUT} = 3.3 \ V$
$C_{BP} = 0.1 \ \mu F$
$I_{LOAD} = 350 \ mA$

Turn-On Delay

$\overline{SD}$ 2 V/div

V_{OUT} 2 V/div

10.00 ms/div

$V_{IN} = 4.2 \ V$
$V_{OUT} = 3.3 \ V$
$C_{BP} = 0.1 \ \mu F$
$I_{LOAD} = 350 \ mA$

Noise Spectrum

$\mu V / \sqrt{Hz}$

$V_{IN} = 4 \ V$
$V_{OUT} = 3$
$I_{OUT} = 150 \ mA$
$C_{IN} = 1 \ \mu F$
$C_{OUT} = 2.2 \ \mu F$
$C_{BP} = 0.1 \ \mu F$

Output Noise

100 μV/div

1 ms/div

$V_{IN} = 4 \ V$
$V_{OUT} = 3 \ V$
$I_{OUT} = 150 \ mA$
$C_{IN} = 1 \ \mu F$
$C_{OUT} = 2.2 \ \mu F$
$C_{BP} = 0.1 \ \mu F$
$BW = 10 \ Hz \ to \ 1 \ MHz$

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TYPICAL WAVEFORMS

Load Transient Response-1
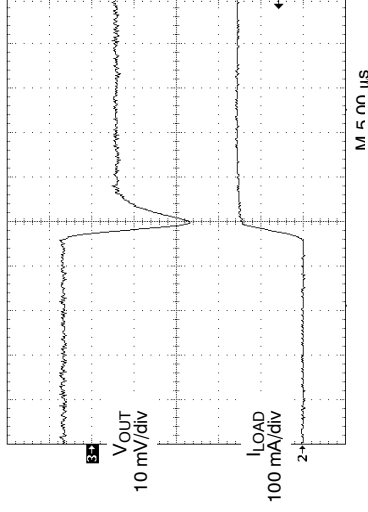
V_{OUT} 10 mV/div
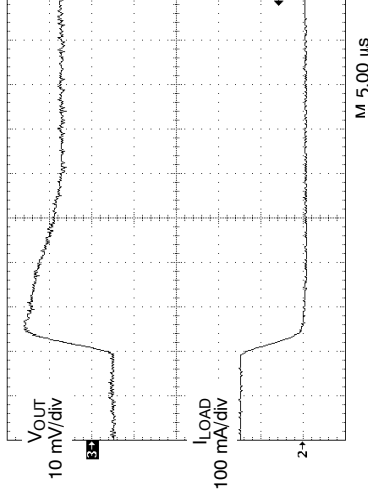
I_{LOAD} 100 mA/div

M 5.00 µs

V_{OUT} = 3.0 V
C_{OUT} = 2.2 µF
I_{LOAD} = 1 to 150 mA
t_{rise} = 2 µsec

Load Transient Response-2

V_{OUT} 10 mV/div

I_{LOAD} 100 mA/div

M 5.00 µs

V_{OUT} = 3.0 V
C_{OUT} = 2.2 µF
I_{LOAD} = 150 to 1 mA
t_{fall} = 2 µsec

Load Transient Response-3

V_{OUT} 10 mV/div

I_{LOAD} 100 mA/div

M 5.00 µs

V_{OUT} = 3.0 V
C_{OUT} = 1.0 µF
I_{LOAD} = 1 to 150 mA
t_{rise} = 2 µsec

Load Transient Response-4

V_{OUT} 10 mV/div

I_{LOAD} 100 mA/div

M 5.00 µs

V_{OUT} = 3.0 V
C_{OUT} = 1.0 µF
I_{LOAD} = 150 to 1 mA
t_{fall} = 2 µsec

Line Transient Response-1

V_{IN} 1 V/div

V_{OUT} 10 mV/div

M 20.0 µs
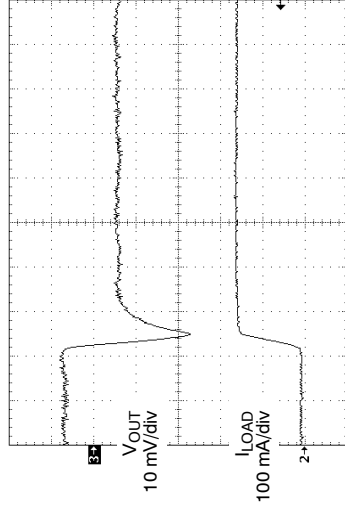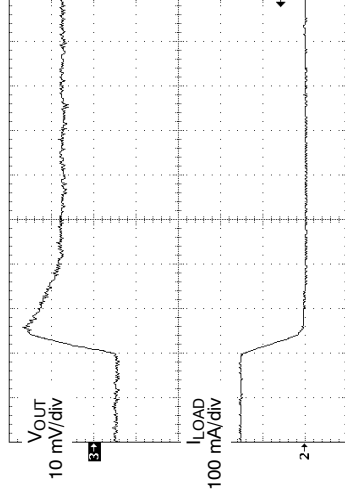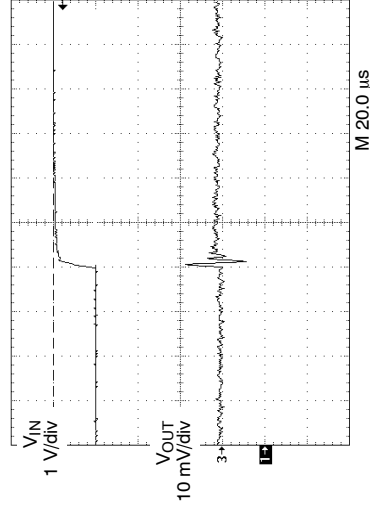
V_{OUT} = 3.0 V
I_{LOAD} = 150 mA
V_{IN} = 4 to 5 V
C_{OUT} = 2.2 µF
t_{rise} = 5 µsec

Line Transient Response-2

V_{IN} 1 V/div

V_{OUT} 10 mV/div

M 20.0 µs
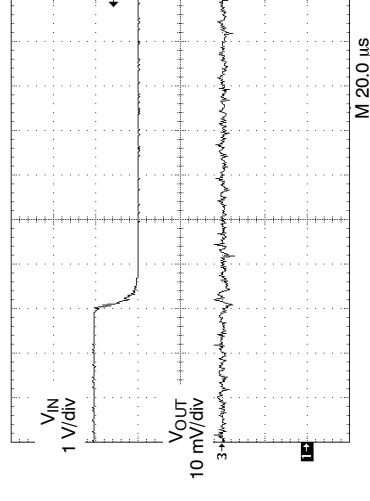
V_{OUT} = 3.0 V
I_{LOAD} = 150 mA
V_{IN} = 5 to 4 V
C_{OUT} = 2.2 µF
t_{fall} = 5 µsec

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BLOCK DIAGRAMS

Switches shown for device in normal operating mode ($\overline{SD}$ = HIGH)
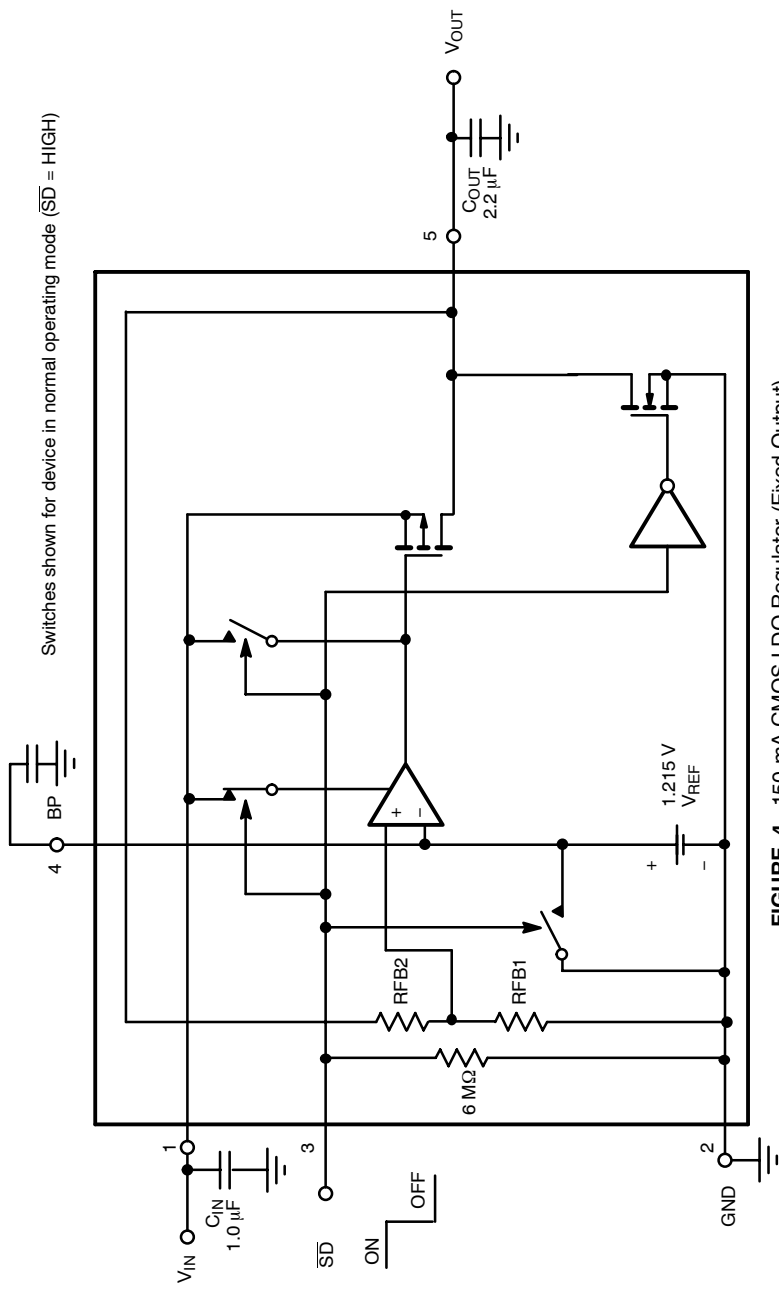
FIGURE 4. 150-mA CMOS LDO Regulator (Fixed Output)
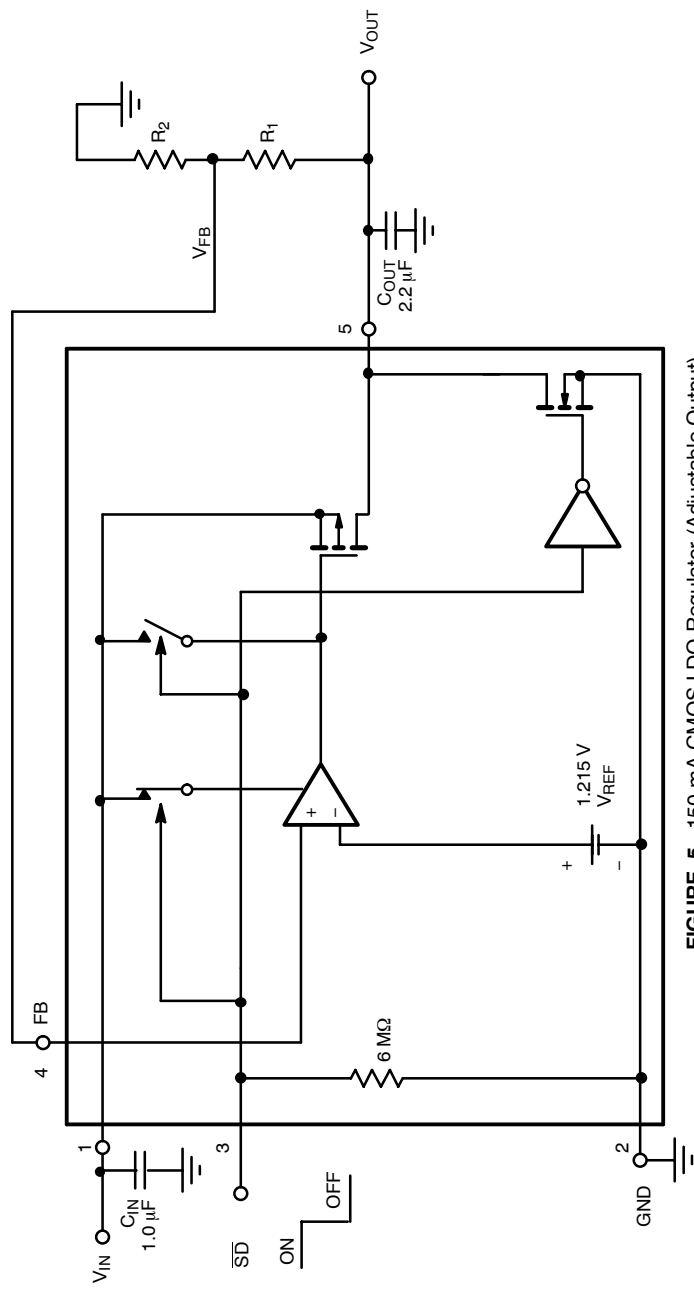
FIGURE 5. 150-mA CMOS LDO Regulator (Adjustable Output)

Si9183

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DETAILED DESCRIPTION

The Si9183 is a low drop out, low quiescent current, linear regulator family with very fast transient response. It is primarily designed for battery powered applications where battery run time is at a premium. The low quiescent current allows extended standby time while low drop out voltage enables the system to fully utilize battery power before recharge. The Si9183 is a very fast regulator with bandwidth exceeding 50 kHz while maintaining low quiescent current at light load conditions. With this bandwidth, the Si9183 is one of the fastest LDO available today. The Si9183 is stable with one of any output capacitor types from 1 µF to 10.0 µF. However, X5R or X7R ceramic capacitors are recommended for best output noise and transient performance.

V_{IN}

V_{IN} is the input supply pin. The bypass capacitor for this pin is not critical as long as the input supply has low enough source impedance. For practical circuits, a 1.0-µF or larger ceramic capacitor is recommended. When the source impedance is not low enough and/or the source is several inches from the Si9183, then a larger input bypass capacitor is needed. It is required that the equivalent impedance (source impedance, wire, and trace impedance in parallel with input bypass capacitor impedance) must be smaller than the input impedance of the Si9183 for stable operation. When the source impedance, wire, and trace impedance are unknown, it is recommended that an input bypass capacitor be used of a value that is equal to or greater than the output capacitor.

V_{OUT}

V_{OUT} is the output voltage of the regulator. Connect a bypass capacitor from V_{OUT} to ground. The output capacitor can be any value from 1.0 µF to 10.0 µF. A ceramic capacitor with X5R or X7R dielectric type is recommended for best output noise, line transient, and load transient performance.

GND

Ground is the common ground connection for V_{IN} and V_{OUT}. It is also the local ground connection for C_{BP}, ADJ, and $\overline{SD}$.

ADJ

For the adjustable output version, use a resistor divider R1 and R2, connect R1 from V_{OUT} to ADJ and R2 from ADJ to ground. R2 should be in the 25-kΩ to 150-kΩ range for low power consumption, while maintaining adequate noise immunity.

The formula below calculates the value of R1, given the desired output voltage and the R2 value,

$$R1 = \frac{\left(V_{OUT} - V_{ADJ}\right)R2}{V_{ADJ}} \qquad (1)$$

V_{ADJ} is nominally 1.215 V.

SHUTDOWN ($\overline{SD}$)

$\overline{SD}$ controls the turning on and off of the Si9183. V_{OUT} is guaranteed to be on when the $\overline{SD}$ pin voltage equals or is greater than 1.2 V. V_{OUT} is guaranteed to be off when the $\overline{SD}$ pin voltage equals or is less than 0.4 V. During shutdown mode, the Si9183 will draw less than 1-µA current from the source. To automatically turn on V_{OUT} whenever the input is applied, tie the $\overline{SD}$ pin to V_{IN}.

C_{BP}

For low noise application, connect a high frequency ceramic capacitor from C_{BP} to ground. A 0.01-µF or a 0.1-µF X5R or X7R is recommended.

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Appendix E-7: LCD Specifications

Appendix E-8: Rectifier Diode Specifications

Surface Mount Silicon Rectifier Diode

NRD Series

FEATURES
- VOLTAGE: 50 TO 1000 VOLTS, CURRENT; 1.0 AMPERE
- CORRESPONDS TO 1N4001 THRU 1N4007 IN SURFACE MOUNT PACKAGE
- FLAT PACK - LOW PROFILE, FOR SURFACE MOUNT APPLICATIONS
- GLASS PASSIVATED CHIP CONSTRUCTION
- HIGH TEMPERATURE SOLDERING (250°C/10 SECONDS)
- EASY PICK AND PLACE

RoHS Compliant
includes all homogeneous materials

*See Part Number System for Details
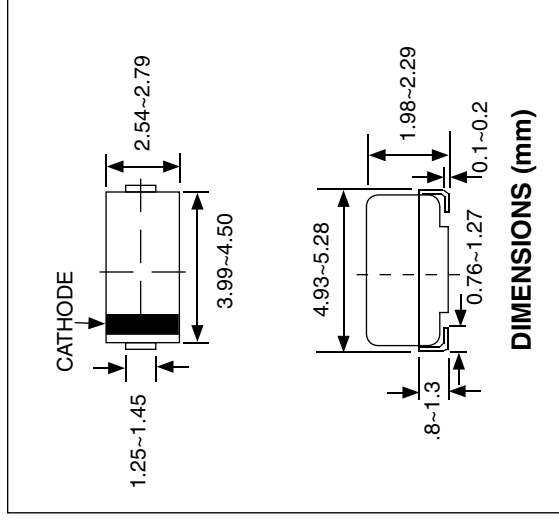
MECHANICAL DATA:
SIZE: SMA/DO-214AC
CASE: Molded epoxy
TERMINALS: Solder plated Copper alloy
POLARITY: Indicated by cathode band
STANDARD PACKAGING: 12mm tape (EIA-RS-481)
WEIGHT: 0.064 gram

PART NUMBERING SYSTEM
NRD 4004 TR -5K F
- Series
- Voltage designator (See table below)
- Tape and Reel
- Reel qty: (5K & 7.5K)
- RoHS Compliant

DIMENSIONS (mm)
CATHODE
2.54~2.79
3.99~4.50
1.25~1.45
1.98~2.29
0.1~0.2
4.93~5.28
0.76~1.27
.8~1.3

MAXIMUM RATINGS (At T$_A$=25°C unless otherwise noted)

Ratings	Symbol	NRD4001	NRD4002	NRD4003	NRD4004	NRD4005	NRD4007	UNITS
Maximum Recurrent Peak Reverse voltage	VRRM	50	100	200	400	600	1000	Volts
Maximum RMS voltage	VRMS	35	70	140	280	420	700	Volts
Maximum DC Blocking Voltage	VDC	50	100	200	400	600	1000	Volts
Maximum Average Forward Rectified Current TA=75°C	Io	1.0						Amps
Peak Forward Surge Current 8.3ms single half sine-wave superimposed on rated load (JEDEC method)	IFSM	30						Amps

ELECTRICAL CHARACTERISTICS (At T$_A$=25°C unless otherwise noted)

Characteristics		Symbol	NRD4001	NRD4002	NRD4003	NRD4004	NRD4005	NRD4007	UNITS
Maximum Forward Voltage at 1.0A DC		VF	1.1						Amps
Maximum Full Load Reverse Current Full Cycle Average @TA=75°C		IR	30						μAmps
Maximum DC Reverse Current at Rated DC Blocking Voltage	@TA=25°C		5.0						μAmps
	@TA=125°C		50						μAmps
Maximum Thermal Resistance (Note 1)		RqJL	30						°C/W
Typical Junction Capacitance (Note 2)		CJ	15						pF
Operating and Storage Temperature Range		TJ, TSTG	-60 ~ +150°C						°C
Maximum Reverse Recovery Time (Note 3)		t$_{rr}$	2.5						μS

NOTES:
1. Thermal resistance junction to terminal, 5mm² (0.013 mm Thick) copper land patterns.
2. Measured at 1.0 MHz and applied average voltage of 4.0VDC.
3. Reverse recovery test conditions: If=0.5A, IR=1.0A, t$_{rr}$=0.25A

RATINGS AND CHARACTERISTICS CURVES
(NRD4001 THRU 4007)

Typical Instanteous Forward Charateristics per Element

Tj=25°C
Pulse Width=300μS
2% Duty Cycle

Instanteous Forward Current (Amps)

Instanteous Forward Voltage (Volts)

Derating Curve for Output Rectified Current

Single Phase Half Wave 60Hz
Resistive or Inductive Load
PCB Mounted
0n 0.315"x0.315" (8.0x8.0mm)
copper Pad Areas

Average Forward Current (Amps)

Temperature (°C)

Maximum Non-repetitive Peak ForwardSurge Current

Forward Surge Current (Amps pk) (Half Sine-wave)

Number of Cycles at 60Hz
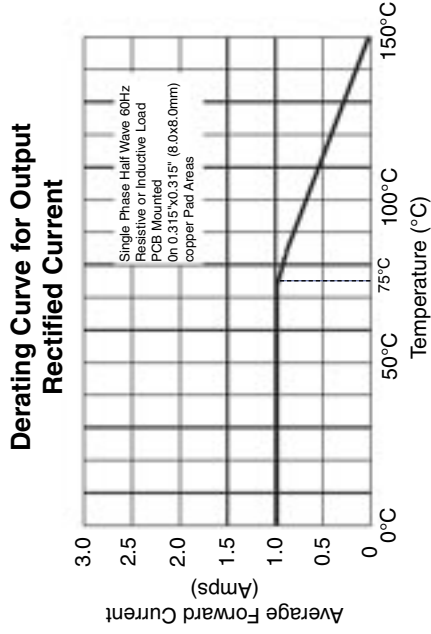
Typical Peak Reverse Characteristics

Tj=100°C
Tj=75°C
Tj=25°C

Instanteous Reverse Current (μA)

Percentage of Peak Reverse Voltage
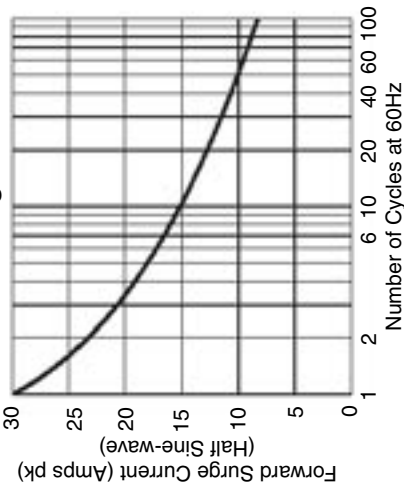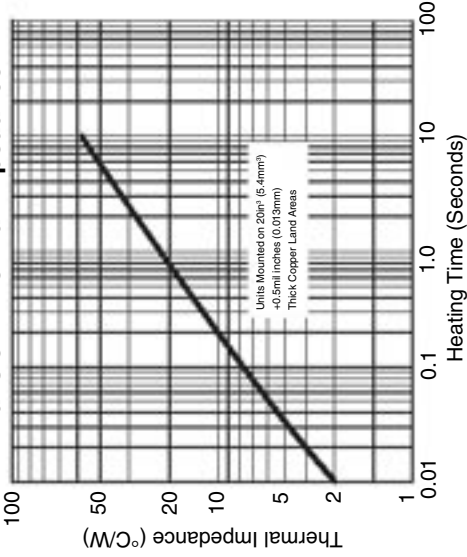
Typcial Junction Capacitance per Element

Tj=25°C
f=1MHz
Vsig=50mVp-p

Capacitance (pF)

Reverse Voltage (Volts)

Transient Thermal Impedance

Units Mounted on 20in² (5.4mm²)
+0.5mil inches (0.013mm)
Thick Copper Land Areas

Thermal Impedance (°C/W)

Heating Time (Seconds)
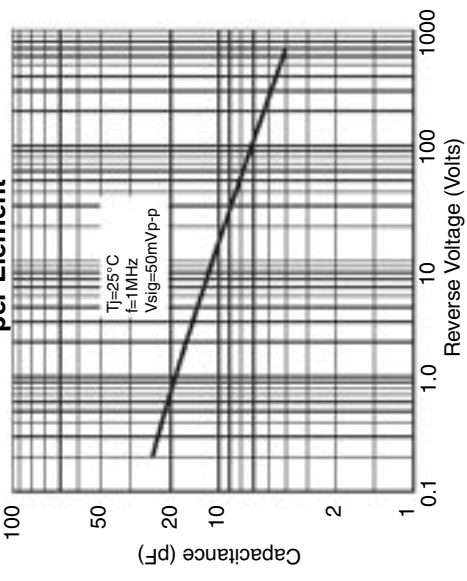
Appendix E-9: Voltage Regulator Specifications

L7800 SERIES

POSITIVE VOLTAGE REGULATORS

- OUTPUT CURRENT TO 1.5A
- OUTPUT VOLTAGES OF 5; 5.2; 6; 8; 8.5; 9; 10; 12; 15; 18; 24V
- THERMAL OVERLOAD PROTECTION
- SHORT CIRCUIT PROTECTION
- OUTPUT TRANSITION SOA PROTECTION

TO-220FP
TO-220FM

TO-3

TO-220

D²PAK

DESCRIPTION

The L7800 series of three-terminal positive regulators is available in TO-220, TO-220FP, TO-220FM, TO-3 and D²PAK packages and several fixed output voltages, making it useful in a wide range of applications. These regulators can provide local on-card regulation, eliminating the distribution problems associated with single point regulation. Each type employs internal current limiting, thermal shut-down and safe area protection, making it essentially indestructible. If adequate heat sinking is provided, they can deliver over 1A output current. Although designed primarily as fixed voltage regulators, these devices can be used with external components to obtain adjustable voltage and currents.
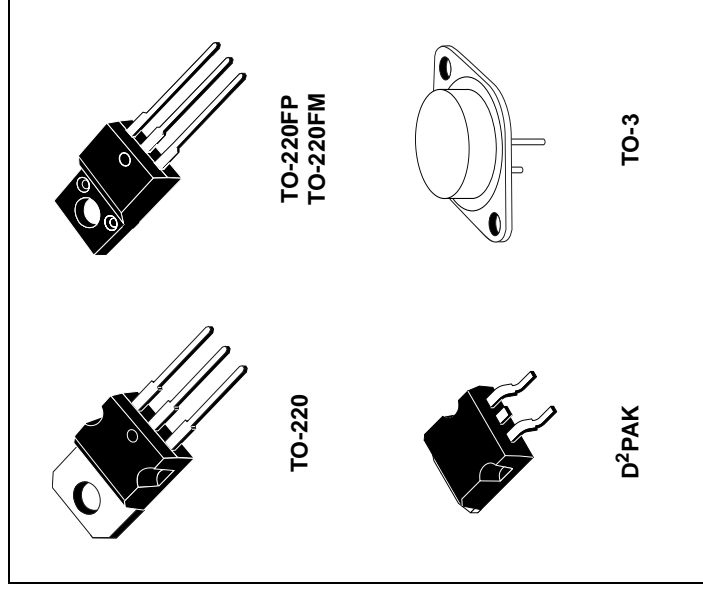
Figure 1: Schematic Diagram

L7800 SERIES
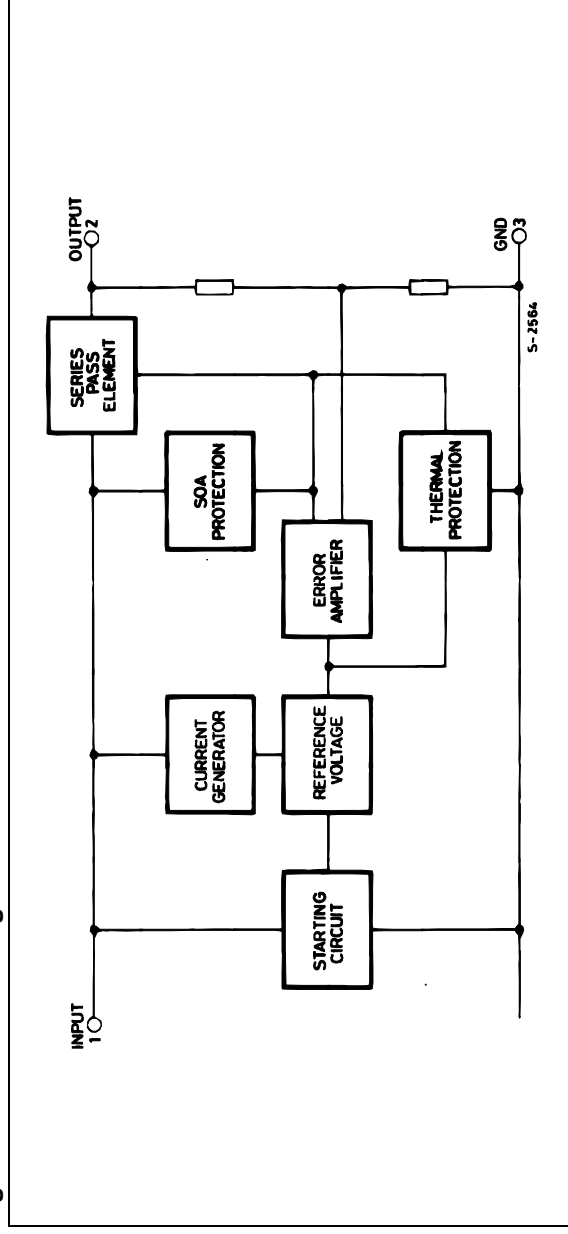
Table 1: Absolute Maximum Ratings

Symbol	Parameter		Value	Unit
V_I	DC Input Voltage	for V_O= 5 to 18V	35	V
		for V_O= 20, 24V	40	
I_O	Output Current		Internally Limited	
P_{tot}	Power Dissipation		Internally Limited	
T_{stg}	Storage Temperature Range		-65 to 150	°C
T_{op}	Operating Junction Temperature Range	for L7800	-55 to 150	°C
		for L7800C	0 to 150	

Absolute Maximum Ratings are those values beyond which damage to the device may occur. Functional operation under these condition is not implied.

Table 2: Thermal Data

Symbol	Parameter	D²PAK	TO-220	TO-220FP	TO-220FM	TO-3	Unit
$R_{thj-case}$	Thermal Resistance Junction-case Max	3	5	5	5	4	°C/W
$R_{thj-amb}$	Thermal Resistance Junction-ambient Max	62.5	50	60	60	35	°C/W

Figure 2: Schematic Diagram
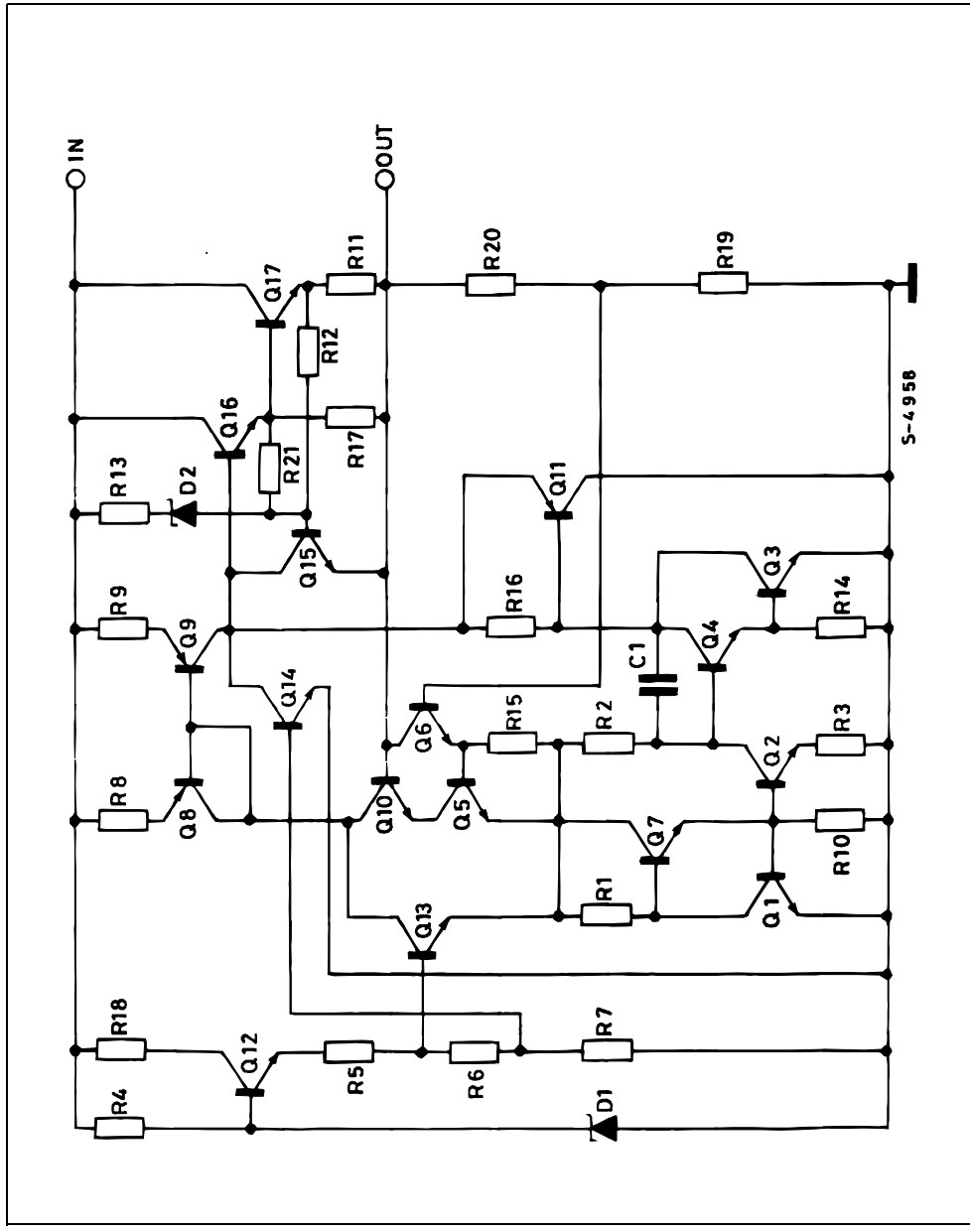
Figure 3: Connection Diagram (top view)

TO-220 (Any Type)

OUTPUT
GROUND
INPUT

CS05710

TO-220FP/TO-220FM

OUTPUT
GROUND
INPUT

CS05700

D²PAK (Any Type)

OUTPUT
INPUT
GND

PC11920

TO-3

OUTPUT
GND
INPUT

CS12750
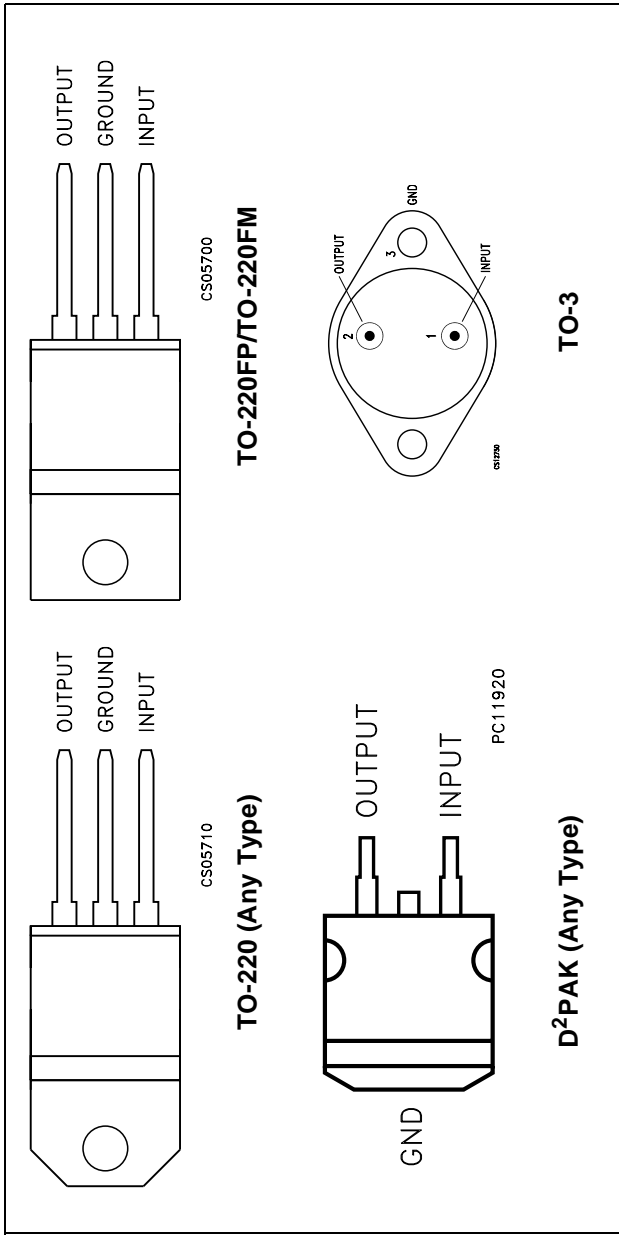
Table 3: Order Codes

TYPE	TO-220 (A Type)	TO-220 (C Type)	TO-220 (E Type)	D²PAK (A Type) (*)	D²PAK (C Type) (T & R)	TO-220FP	TO-220FM	TO-3
L7805								L7805T
L7805C	L7805CV	L7805C-V	L7805CV1	L7805CD2T	L7805C-D2TR	L7805CP	L7805CF	L7805CT
L7852C	L7852CV	L7852C		L7852CD2T		L7852CP	L7852CF	L7852CT
L7806								L7806T
L7806C	L7806CV	L7806C-V		L7806CD2T		L7806CP	L7806CF	L7806CT
L7808								L7808T
L7808C	L7808CV	L7808C-V		L7808CD2T		L7808CP	L7808CF	L7808CT
L7885C	L7885CV	L7885C-V		L7885CD2T		L7885CP	L7885CF	L7885CT
L7809C	L7809CV	L7809C-V		L7809CD2T		L7809CP	L7809CF	L7809CT
L7810C	L7810CV			L7810CD2T		L7810CP		
L7812								L7812T
L7812C	L7812CV	L7812C-V		L7812CD2T		L7812CP	L7812CF	L7812CT
L7815								L7815T
L7815C	L7815CV	L7815C-V		L7815CD2T		L7815CP	L7815CF	L7815CT
L7818								L7818T
L7818C	L7818CV			L7818CD2T		L7818CP	L7818CF	L7818CT
L7820								L7820T
L7820C	L7820CV			L7820CD2T		L7820CP	L7820CF	L7820CT
L7824								L7824T
L7824C	L7824CV			L7824CD2T		L7824CP	L7824CF	L7824CT

(*) Available in Tape & Reel with the suffix "-TR".

Figure 4: Application Circuits
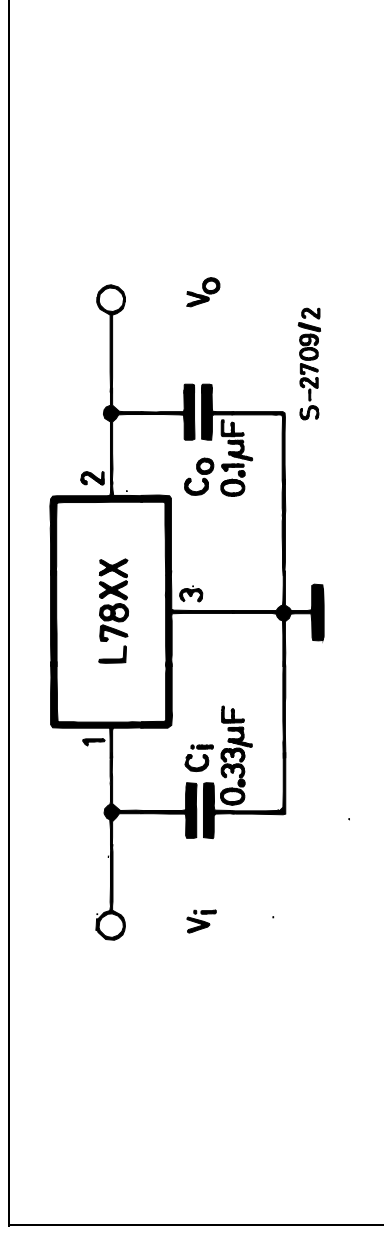
TEST CIRCUITS

Figure 5: DC Parameter
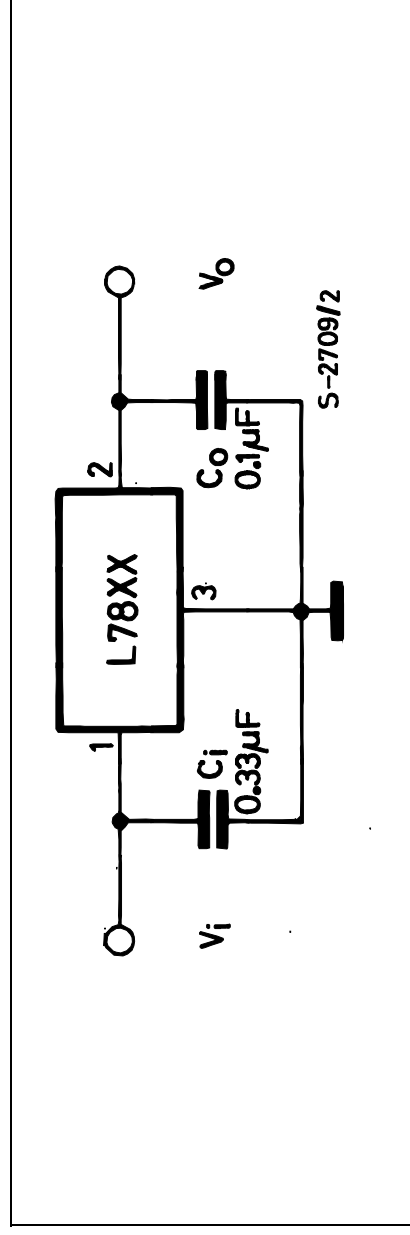
Figure 6: Load Regulation
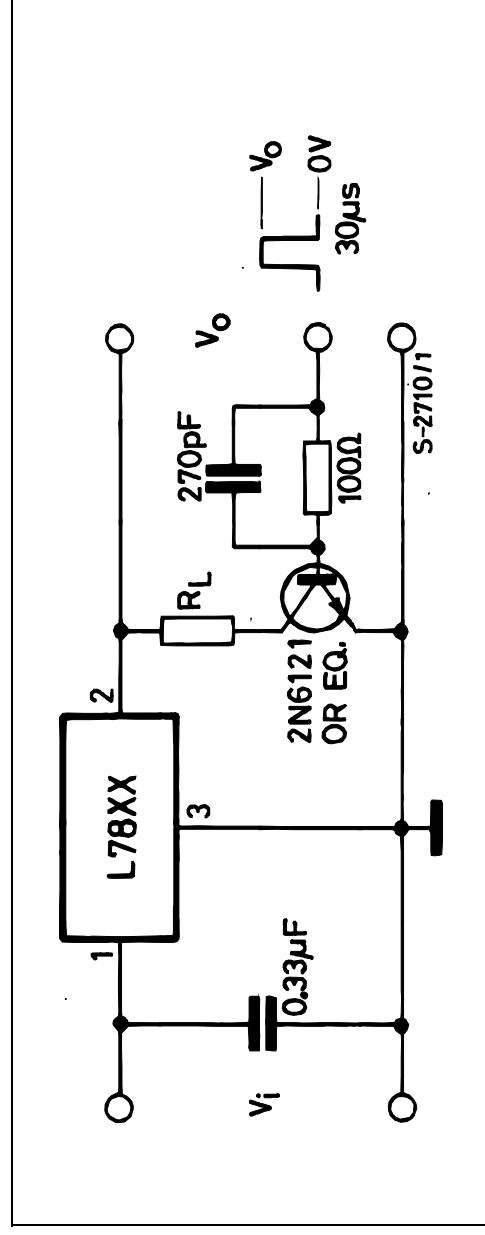
Figure 7: Ripple Rejection
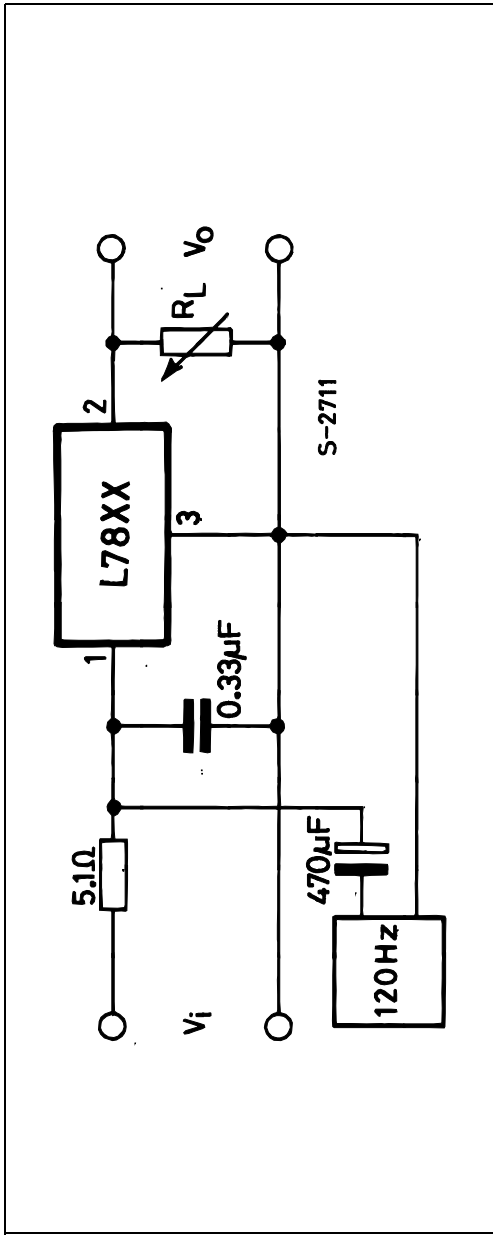
Table 4: Electrical Characteristics Of L7805 (refer to the test circuits, T_J = -55 to 150°C, V_I = 10V, I_O = 500 mA, C_I = 0.33 μF, C_O = 0.1 μF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		4.8	5	5.2	V
V_O	Output Voltage	I_O = 5 mA to 1 A $\quad P_O \leq 15W$ V_I = 8 to 20 V		4.65	5	5.35	V
ΔV_O(*)	Line Regulation	V_I = 7 to 25 V	T_J = 25°C		3	50	mV
		V_I = 8 to 12 V	T_J = 25°C		1	25	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			100	mV
		I_O = 250 to 750 mA	T_J = 25°C			25	
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 8 to 25 V				0.8	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			0.6		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz $\quad T_J$ = 25°C				40	μV/V_O
SVR	Supply Voltage Rejection	V_I = 8 to 18 V $\qquad$ f = 120Hz		68			dB
V_d	Dropout Voltage	I_O = 1 A $\quad T_J$ = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			17		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V $\quad T_J$ = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 5: Electrical Characteristics Of L7806 (refer to the test circuits, T_J = -55 to 150°C, V_I = 11V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		5.75	6	6.25	V
V_O	Output Voltage	I_O = 5 mA to 1 A, V_I = 9 to 21 V	$P_O \leq 15W$	5.65	6	6.35	V
ΔV_O(*)	Line Regulation	V_I = 8 to 25 V	T_J = 25°C			60	mV
		V_I = 9 to 13 V	T_J = 25°C			30	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			100	mV
		I_O = 250 to 750 mA	T_J = 25°C			30	mV
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 9 to 25 V				0.8	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			0.7		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 9 to 19 V	f = 120Hz	65			dB
V_d	Dropout Voltage	I_O = 1 A	T_J = 25°C		2	2.5	V
R_O	Output Resistance	f = 1 KHz			19		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V	T_J = 25°C		0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 6: Electrical Characteristics Of L7808 (refer to the test circuits, T_J = -55 to 150°C, V_I = 14V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		7.7	8	8.3	V
V_O	Output Voltage	I_O = 5 mA to 1 A, V_I = 11.5 to 23 V	$P_O \leq 15W$	7.6	8	8.4	V
ΔV_O(*)	Line Regulation	V_I = 10.5 to 25 V	T_J = 25°C			80	mV
		V_I = 11 to 17 V	T_J = 25°C			40	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			100	mV
		I_O = 250 to 750 mA	T_J = 25°C			40	mV
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 11.5 to 25 V				0.8	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 11.5 to 21.5 V	f = 120Hz	62			dB
V_d	Dropout Voltage	I_O = 1 A	T_J = 25°C		2	2.5	V
R_O	Output Resistance	f = 1 KHz			16		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V	T_J = 25°C		0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 7: Electrical Characteristics Of L7812 (refer to the test circuits, T_J = -55 to 150°C, V_I = 19V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		11.5	12	12.5	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$	V_I = 15.5 to 27 V	11.4	12	12.6	V
ΔV_O(*)	Line Regulation	V_I = 14.5 to 30 V	T_J = 25°C			120	mV
		V_I = 16 to 22 V	T_J = 25°C			60	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			100	mV
		I_O = 250 to 750 mA	T_J = 25°C			60	mV
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 15 to 30 V				0.8	mA
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			1.5		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 15 to 25 V	f = 120Hz	61			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			18		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 8: Electrical Characteristics Of L7815 (refer to the test circuits, T_J = -55 to 150°C, V_I = 23V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		14.4	15	15.6	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$ V_I = 18.5 to 30 V		14.25	15	15.75	V
ΔV_O(*)	Line Regulation	V_I = 17.5 to 30 V	T_J = 25°C			150	mV
		V_I = 20 to 26 V	T_J = 25°C			75	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			150	mV
		I_O = 250 to 750 mA	T_J = 25°C			75	mV
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 18.5 to 30 V				0.8	mA
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			1.8		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 18.5 to 28.5 V	f = 120Hz	60			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			19		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 9: Electrical Characteristics Of L7818 (refer to the test circuits, T_J = -55 to 150°C, V_I = 26V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		17.3	18	18.7	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le$ 15W		17.1	18	18.9	V
		V_I = 22 to 33 V					
ΔV_O(*)	Line Regulation	V_I = 21 to 33 V	T_J = 25°C			180	mV
		V_I = 24 to 30 V	T_J = 25°C			90	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			180	mV
		I_O = 250 to 750 mA	T_J = 25°C			90	
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 22 to 33 V				0.8	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			2.3		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 22 to 32 V	f = 120Hz	59			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			22		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 10: Electrical Characteristics Of L7820 (refer to the test circuits, T_J = -55 to 150°C, V_I = 28V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		19.2	20	20.8	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le$ 15W		19	20	21	V
		V_I = 24 to 35 V					
ΔV_O(*)	Line Regulation	V_I = 22.5 to 35 V	T_J = 25°C			200	mV
		V_I = 26 to 32 V	T_J = 25°C			100	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			200	mV
		I_O = 250 to 750 mA	T_J = 25°C			100	
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 24 to 35 V				0.8	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			2.5		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 24 to 35 V	f = 120Hz	58			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			24		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 11: Electrical Characteristics Of L7824 (refer to the test circuits, T_J = -55 to 150°C, V_I = 33V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		23	24	25	V
V_O	Output Voltage	I_O = 5 mA to 1 A, V_I = 28 to 38 V	$P_O \leq 15W$	22.8	24	25.2	V
ΔV_O(*)	Line Regulation	V_I = 27 to 38 V	T_J = 25°C			240	mV
		V_I = 30 to 36 V	T_J = 25°C			120	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			240	mV
		I_O = 250 to 750 mA	T_J = 25°C			120	mV
I_d	Quiescent Current	T_J = 25°C				6	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 28 to 38 V				0.8	mA
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			3		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C			40	µV/V_O
SVR	Supply Voltage Rejection	V_I = 28 to 38 V	f = 120Hz	56			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2	2.5	V
R_O	Output Resistance	f = 1 KHz			28		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75	1.2	A
I_{scp}	Short Circuit Peak Current	T_J = 25°C		1.3	2.2	3.3	A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 12: Electrical Characteristics Of L7805C (refer to the test circuits, T_J = 0 to 125°C, V_I = 10V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		4.8	5	5.2	V
V_O	Output Voltage	I_O = 5 mA to 1 A, V_I = 7 to 20 V	$P_O \leq 15W$	4.75	5	5.25	V
ΔV_O(*)	Line Regulation	V_I = 7 to 25 V	T_J = 25°C		3	100	mV
		V_I = 8 to 12 V	T_J = 25°C		1	50	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			100	mV
		I_O = 250 to 750 mA	T_J = 25°C			50	mV
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 7 to 25 V				0.8	mA
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1.1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		40		µV/V_O
SVR	Supply Voltage Rejection	V_I = 8 to 18 V	f = 120Hz	62			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			17		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 13: Electrical Characteristics Of L7852C (refer to the test circuits, T_J = 0 to 125°C, V_I = 10V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		5.0	5.2	5.4	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \leq$ 15W V_I = 8 to 20 V		4.95	5.2	5.45	V
$\Delta V_O(*)$	Line Regulation	V_I = 7 to 25 V	T_J = 25°C		3	105	mV
		V_I = 8 to 12 V	T_J = 25°C		1	52	
$\Delta V_O(*)$	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			105	mV
		I_O = 250 to 750 mA	T_J = 25°C			52	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 7 to 25 V				1.3	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		42		µV/V_O
SVR	Supply Voltage Rejection	V_I = 8 to 18 V	f = 120Hz	61			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			17		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.75		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 14: Electrical Characteristics Of L7806C (refer to the test circuits, T_J = 0 to 125°C, V_I = 11V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		5.75	6	6.25	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \leq$ 15W V_I = 8 to 21 V		5.7	6	6.3	V
$\Delta V_O(*)$	Line Regulation	V_I = 8 to 25 V	T_J = 25°C			120	mV
		V_I = 9 to 13 V	T_J = 25°C			60	
$\Delta V_O(*)$	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			120	mV
		I_O = 250 to 750 mA	T_J = 25°C			60	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 8 to 25 V				1.3	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-0.8		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		45		µV/V_O
SVR	Supply Voltage Rejection	V_I = 9 to 19 V	f = 120Hz	59			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			19		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.55		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 15: Electrical Characteristics Of L7808C (refer to the test circuits, T_J = 0 to 125°C, V_I = 14V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		7.7	8	8.3	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$		7.6	8	8.4	V
		V_I = 10.5 to 25 V					
ΔV_O(*)	Line Regulation	V_I = 10.5 to 25 V	T_J = 25°C			160	mV
		V_I = 11 to 17 V	T_J = 25°C			80	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			160	mV
		I_O = 250 to 750 mA	T_J = 25°C			80	mV
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 10.5 to 25 V				1	
$\Delta V_O / \Delta T$	Output Voltage Drift	I_O = 5 mA			-0.8		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		52		µV/V_O
SVR	Supply Voltage Rejection	V_I = 11.5 to 21.5 V	f = 120Hz	56			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			16		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.45		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 16: Electrical Characteristics Of L7885C (refer to the test circuits, T_J = 0 to 125°C, V_I = 14.5V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		8.2	8.5	8.8	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$		8.1	8.5	8.9	V
		V_I = 11 to 26 V					
ΔV_O(*)	Line Regulation	V_I = 11 to 27 V	T_J = 25°C			160	mV
		V_I = 11.5 to 17.5 V	T_J = 25°C			80	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			160	mV
		I_O = 250 to 750 mA	T_J = 25°C			80	mV
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 11 to 27 V				1	
$\Delta V_O / \Delta T$	Output Voltage Drift	I_O = 5 mA			-0.8		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		55		µV/V_O
SVR	Supply Voltage Rejection	V_I = 12 to 22 V	f = 120Hz	56			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			16		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.45		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 17: Electrical Characteristics Of L7809C (refer to the test circuits, T_J = 0 to 125°C, V_I = 15V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

| Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		8.64	9	9.36	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$ V_I = 11.5 to 26 V		8.55	9	9.45	V
ΔV_O(*)	Line Regulation	V_I = 11.5 to 26 V	T_J = 25°C			180	mV
		V_I = 12 to 18 V	T_J = 25°C			90	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			180	mV
		I_O = 250 to 750 mA	T_J = 25°C			90	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 11.5 to 26 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		70		µV/V_O
SVR	Supply Voltage Rejection	V_I = 12 to 23 V f = 120Hz		55			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			17		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.40		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 18: Electrical Characteristics Of L7810C (refer to the test circuits, T_J = 0 to 125°C, V_I = 16V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		9.6	10	10.4	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \le 15W$ V_I = 12.5 to 26 V		9.5	10	10.5	V
ΔV_O(*)	Line Regulation	V_I = 12.5 to 26 V	T_J = 25°C			200	mV
		V_I = 13.5 to 19 V	T_J = 25°C			100	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			200	mV
		I_O = 250 to 750 mA	T_J = 25°C			100	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 12.5 to 26 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		70		µV/V_O
SVR	Supply Voltage Rejection	V_I = 13 to 23 V f = 120Hz		55			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			17		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.40		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 19: Electrical Characteristics Of L7812C (refer to the test circuits, T_J = 0 to 125°C, V_I = 19V, I_O = 500 mA, C_I = 0.33 μF, C_O = 0.1 μF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		11.5	12	12.5	V
V_O	Output Voltage	I_O = 5 mA to 1 A $\quad$ P_O ≤ 15W, V_I = 14.5 to 27 V		11.4	12	12.6	V
ΔV_O(*)	Line Regulation	V_I = 14.5 to 30 V	T_J = 25°C			240	mV
		V_I = 16 to 22 V	T_J = 25°C			120	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			240	mV
		I_O = 250 to 750 mA	T_J = 25°C			120	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 14.5 to 30 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		75		μV/V_O
SVR	Supply Voltage Rejection	V_I = 15 to 25 V $\quad$ f = 120Hz		55			dB
V_d	Dropout Voltage	I_O = 1 A $\quad$ T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			18		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V $\quad$ T_J = 25°C			0.35		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 20: Electrical Characteristics Of L7815C (refer to the test circuits, T_J = 0 to 125°C, V_I = 23V, I_O = 500 mA, C_I = 0.33 μF, C_O = 0.1 μF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		14.5	15	15.6	V
V_O	Output Voltage	I_O = 5 mA to 1 A $\quad$ P_O ≤ 15W, V_I = 17.5 to 30 V		14.25	15	15.75	V
ΔV_O(*)	Line Regulation	V_I = 17.5 to 30 V	T_J = 25°C			300	mV
		V_I = 20 to 26 V	T_J = 25°C			150	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			300	mV
		I_O = 250 to 750 mA	T_J = 25°C			150	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 17.5 to 30 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		90		μV/V_O
SVR	Supply Voltage Rejection	V_I = 18.5 to 28.5 V $\quad$ f = 120Hz		54			dB
V_d	Dropout Voltage	I_O = 1 A $\quad$ T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			19		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V $\quad$ T_J = 25°C			0.23		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.2		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 21: Electrical Characteristics Of L7818C (refer to the test circuits, T_J = 0 to 125°C, V_I = 26V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		17.3	18	18.7	V
V_O	Output Voltage	I_O = 5 mA to 1 A $\quad P_O \leq$ 15W V_I = 21 to 33 V		17.1	18	18.9	V
ΔV_O(*)	Line Regulation	V_I = 21 to 33 V	T_J = 25°C			360	mV
		V_I = 24 to 30 V	T_J = 25°C			180	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			360	mV
		I_O = 250 to 750 mA	T_J = 25°C			180	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 21 to 33 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		110		µV/V_O
SVR	Supply Voltage Rejection	V_I = 22 to 32 V	f = 120Hz	53			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			22		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.20		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.1		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 22: Electrical Characteristics Of L7820C (refer to the test circuits, T_J = 0 to 125°C, V_I = 28V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		19.2	20	20.8	V
V_O	Output Voltage	I_O = 5 mA to 1 A $\quad P_O \leq$ 15W V_I = 23 to 35 V		19	20	21	V
ΔV_O(*)	Line Regulation	V_I = 22.5 to 35 V	T_J = 25°C			400	mV
		V_I = 26 to 32 V	T_J = 25°C			200	
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			400	mV
		I_O = 250 to 750 mA	T_J = 25°C			200	
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 23 to 35 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		150		µV/V_O
SVR	Supply Voltage Rejection	V_I = 24 to 35 V	f = 120Hz	52			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			24		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.18		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.1		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Table 23: Electrical Characteristics Of L7824C (refer to the test circuits, T_J = 0 to 125°C, V_I = 33V, I_O = 500 mA, C_I = 0.33 µF, C_O = 0.1 µF unless otherwise specified).

Symbol	Parameter	Test Conditions		Min.	Typ.	Max.	Unit
V_O	Output Voltage	T_J = 25°C		23	24	25	V
V_O	Output Voltage	I_O = 5 mA to 1 A $P_O \leq$ 15W V_I = 27 to 38 V		22.8	24	25.2	V
ΔV_O(*)	Line Regulation	V_I = 27 to 38 V	T_J = 25°C			480	mV
		V_I = 30 to 36 V	T_J = 25°C			240	mV
ΔV_O(*)	Load Regulation	I_O = 5 mA to 1.5 A	T_J = 25°C			480	mV
		I_O = 250 to 750 mA	T_J = 25°C			240	mV
I_d	Quiescent Current	T_J = 25°C				8	mA
ΔI_d	Quiescent Current Change	I_O = 5 mA to 1 A				0.5	mA
		V_I = 27 to 38 V				1	
$\Delta V_O/\Delta T$	Output Voltage Drift	I_O = 5 mA			-1.5		mV/°C
eN	Output Noise Voltage	B =10Hz to 100KHz	T_J = 25°C		170		µV/V_O
SVR	Supply Voltage Rejection	V_I = 28 to 38 V f = 120Hz		50			dB
V_d	Dropout Voltage	I_O = 1 A T_J = 25°C			2		V
R_O	Output Resistance	f = 1 KHz			28		mΩ
I_{sc}	Short Circuit Current	V_I = 35 V T_J = 25°C			0.15		A
I_{scp}	Short Circuit Peak Current	T_J = 25°C			2.1		A

(*) Load and line regulation are specified at constant junction temperature. Changes in V_O due to heating effects must be taken into account separately. Pulse testing with low duty cycle is used.

Figure 8: Dropout Voltage vs Junction Temperature
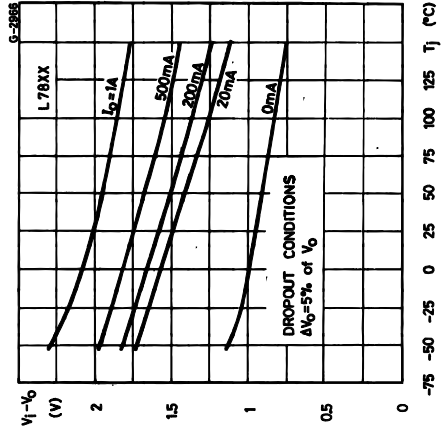
Figure 9: Peak Output Current vs Input/output Differential Voltage
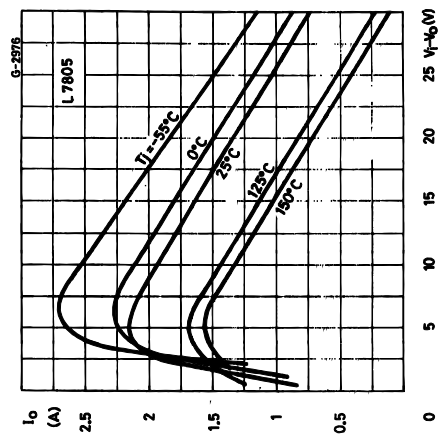
Figure 10: Supply Voltage Rejection vs Frequency

Figure 11: Output Voltage vs Junction Temperature

Figure 12: Output Impedance vs Frequency

Figure 13: Quiescent Current vs Junction Temperature

Figure 14: Load Transient Response
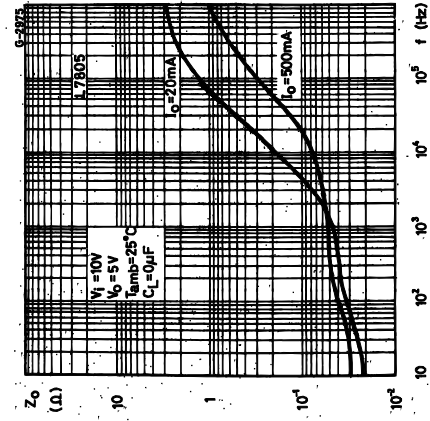
Figure 15: Line Transient Response
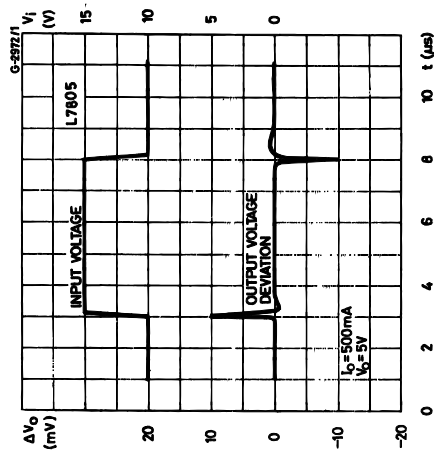
Figure 16: Quiescent Current vs Input Voltage
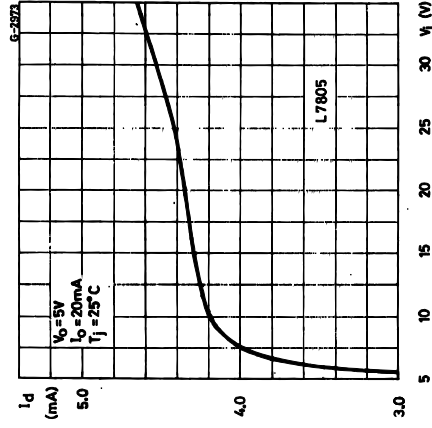
Figure 17: Fixed Output Regulator
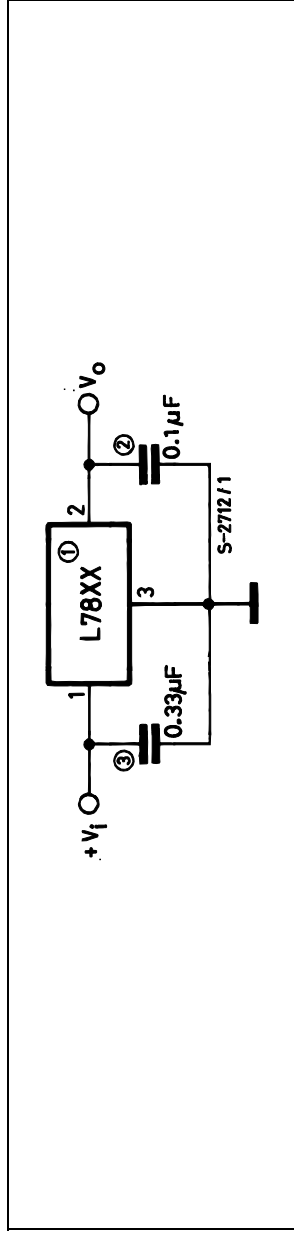
NOTE:
1. To specify an output voltage, substitute voltage value for "XX".
2. Although no output capacitor is need for stability, it does improve transient response.
3. Required if regulator is locate an appreciable distance from power supply filter.

Figure 18: Current Regulator

$$I_O = \frac{V_{XX}}{R_1} + I_d$$
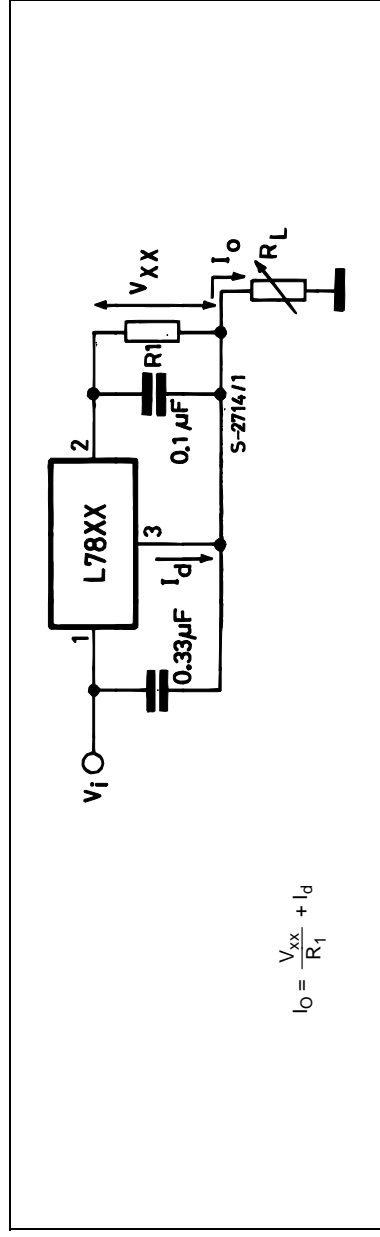
Figure 19: Circuit for Increasing Output Voltage

$I_{R1} \geq 5 \; I_d$

$$V_O = V_{XX}(1 + \frac{R_2}{R_1}) + I_d R_2$$
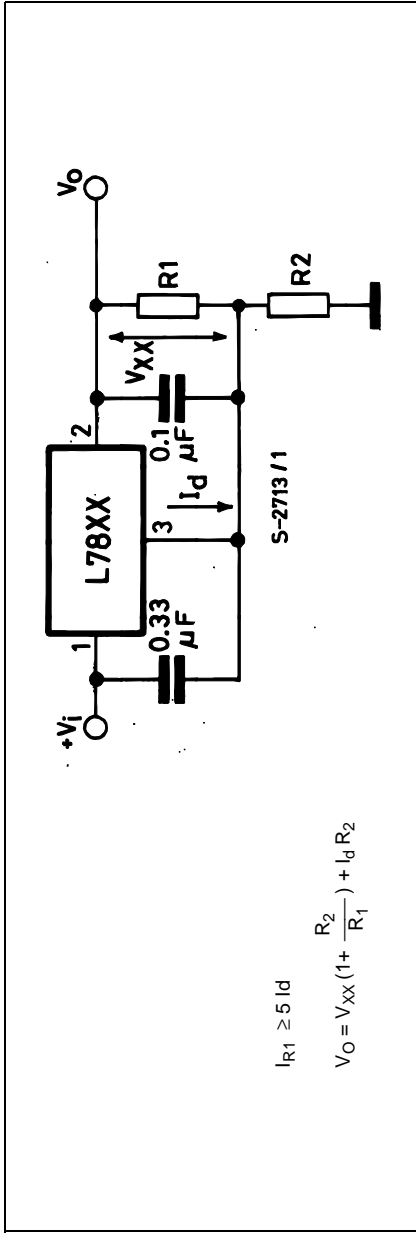
Figure 20: Adjustable Output Regulator (7 to 30V)
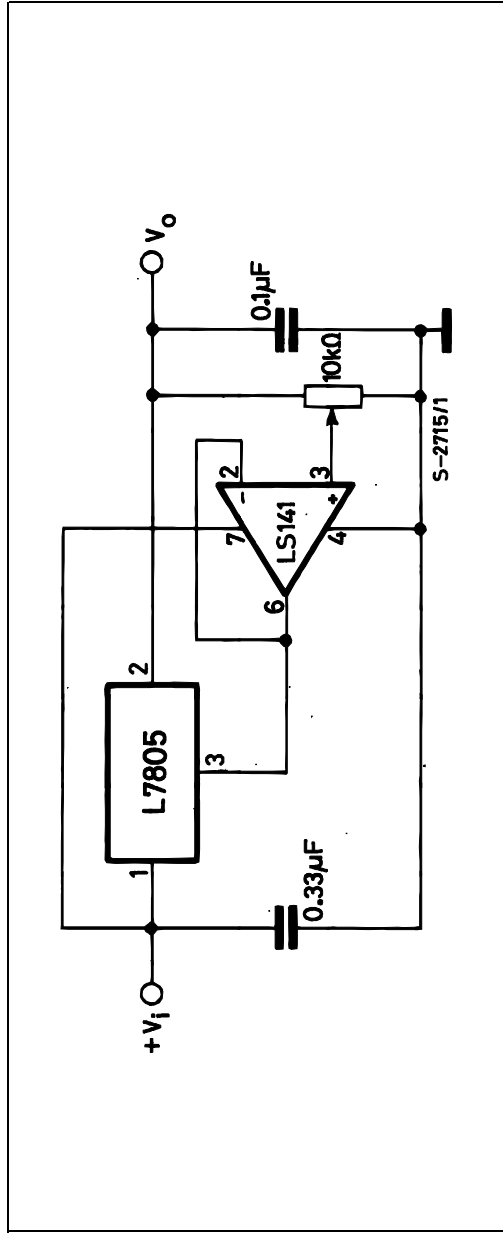
Figure 21: 0.5 to 10V Regulator

$$V_O = V_{xx} \frac{R_4}{R_1}$$
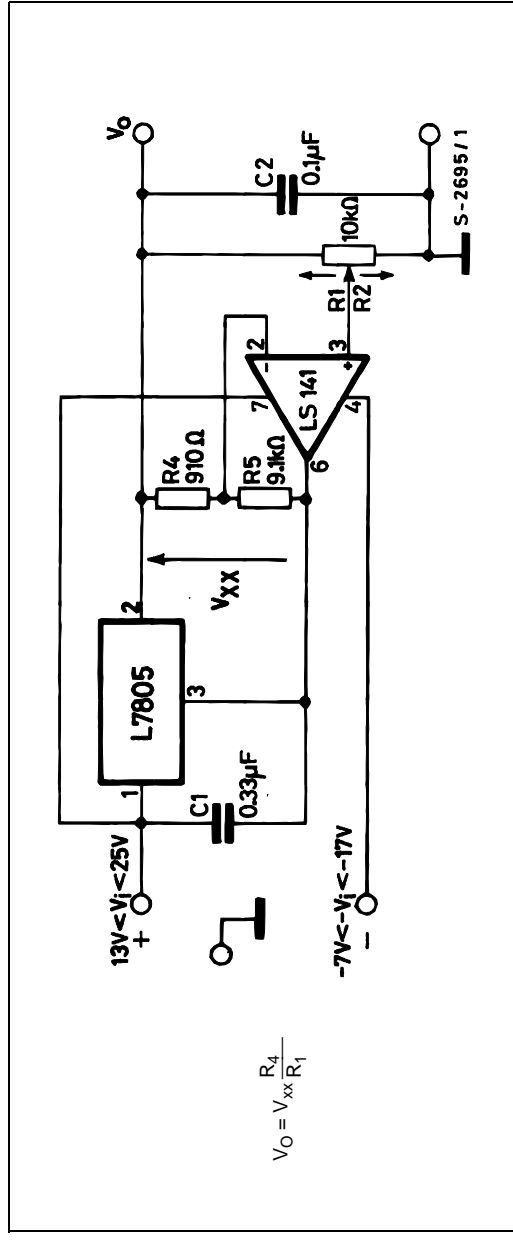
Figure 22: High Current Voltage Regulator

$$R_1 = \frac{V_{BEQ1}}{I_{REQ} - \frac{I_{Q1}}{\beta_{Q1}}}$$

$$I_O = I_{REG} + Q_1 \left(I_{REG} - \frac{V_{BEQ1}}{R_1} \right)$$
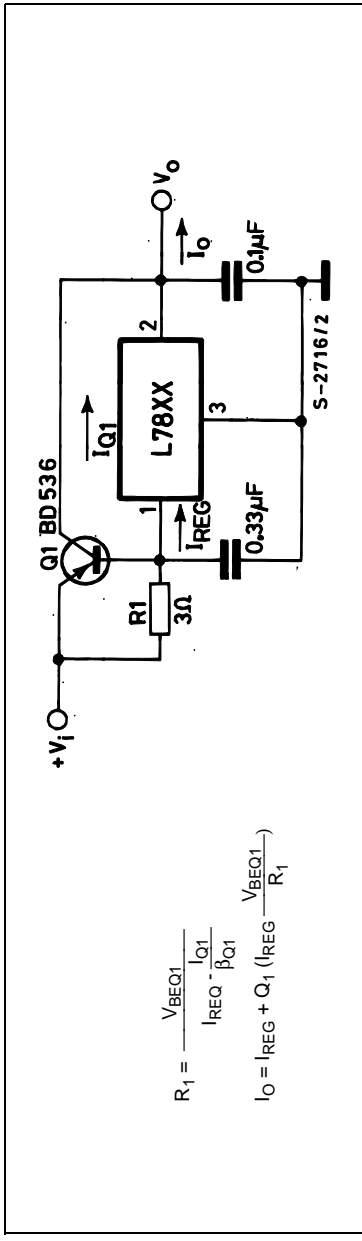
Figure 23: High Output Current with Short Circuit Protection

Q1 = BD 534
Q2 = 2N6124

$$R_{SC} = \frac{V_{BEQ2}}{I_{SC}}$$
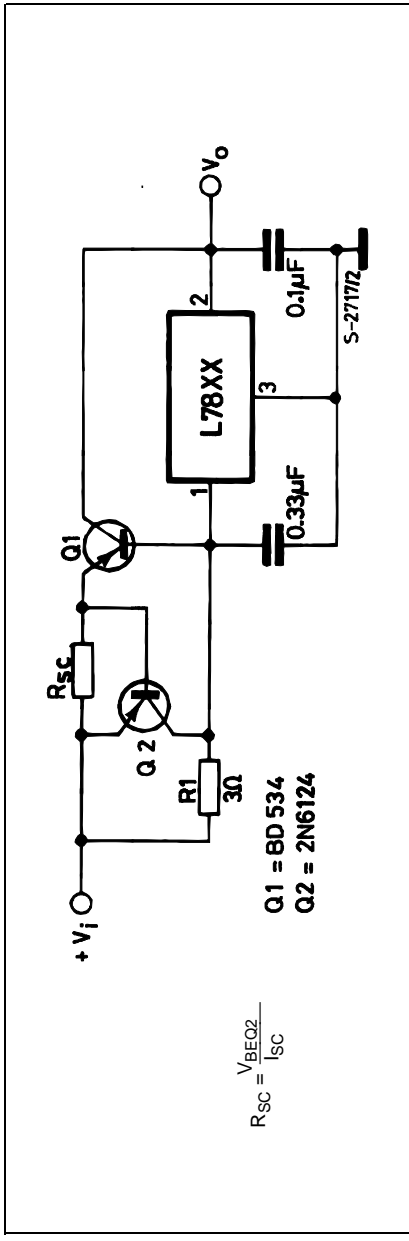
Figure 24: Tracking Voltage Regulator
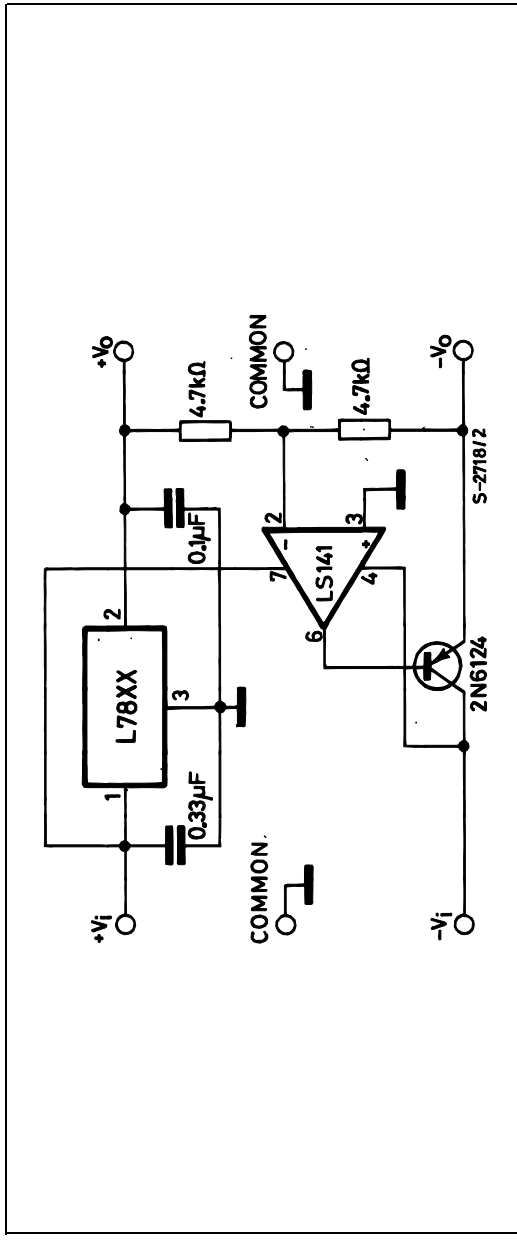
Figure 25: Split Power Supply (± 15V - 1 A)
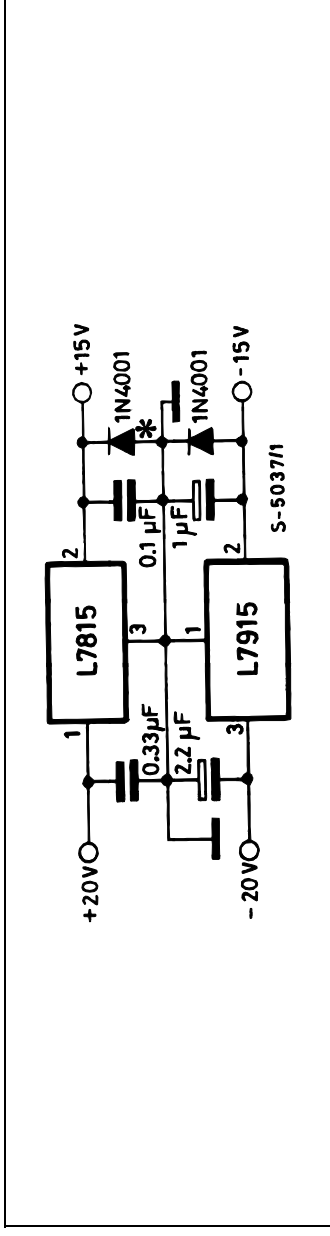
* Against potential latch-up problems.

Figure 26: Negative Output Voltage Circuit
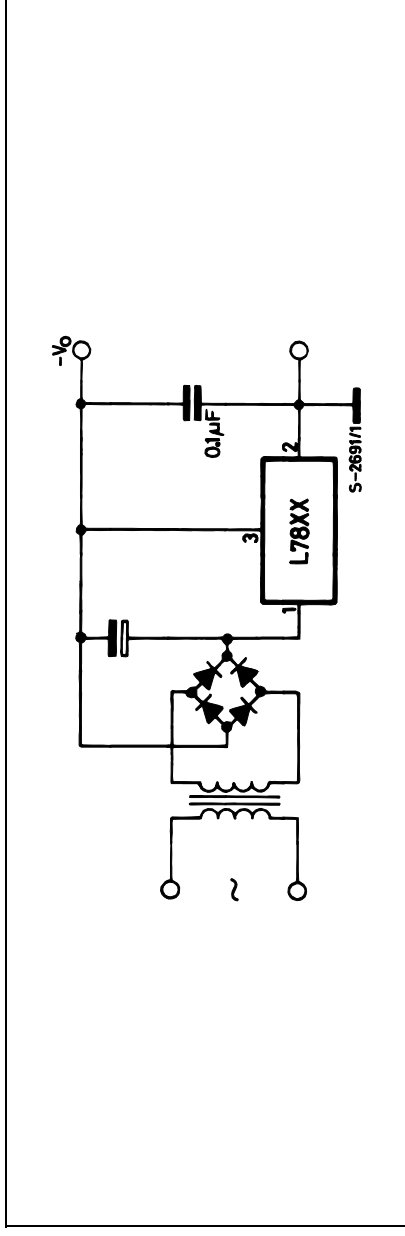
Figure 27: Switching Regulator
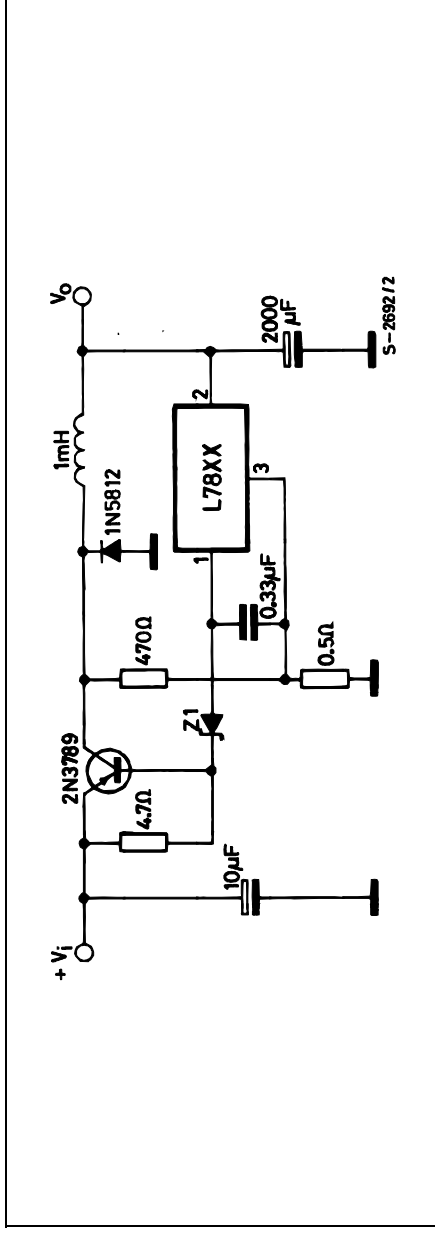
Figure 28: High Input Voltage Circuit

$$V_{IN} = V_I - (V_Z + V_{BE})$$
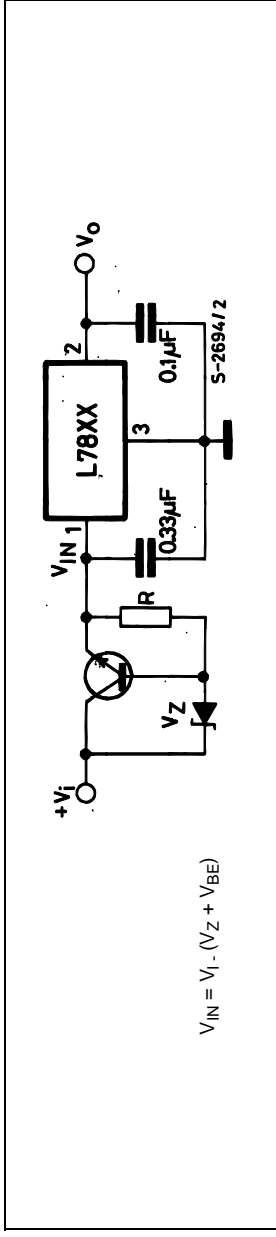
Figure 29: High Input Voltage Circuit
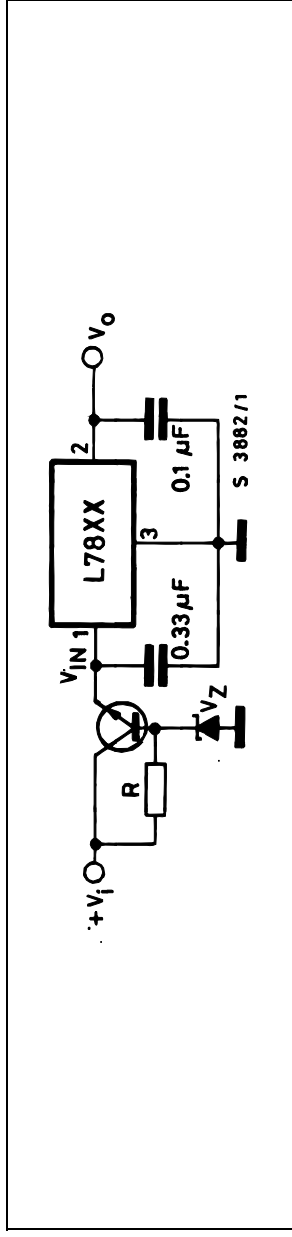
Figure 30: High Output Voltage Regulator
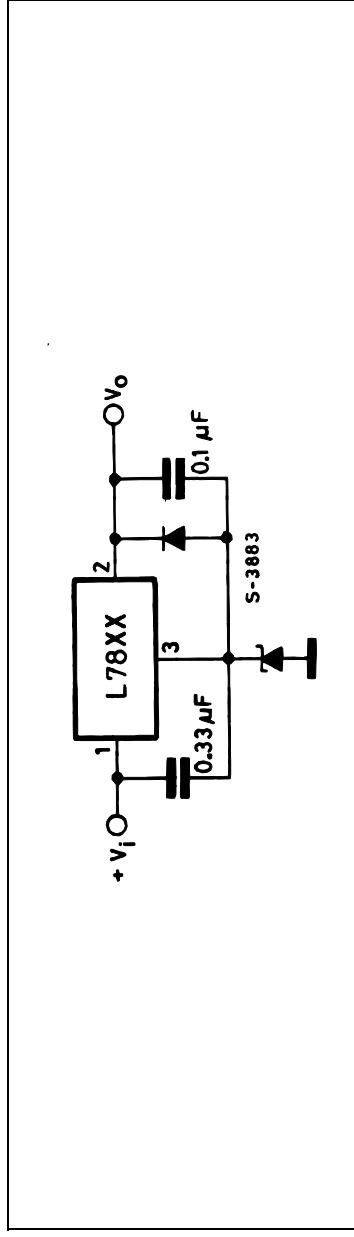
Figure 31: High Input and Output Voltage

$$V_O = V_{XX} + V_{Z1}$$
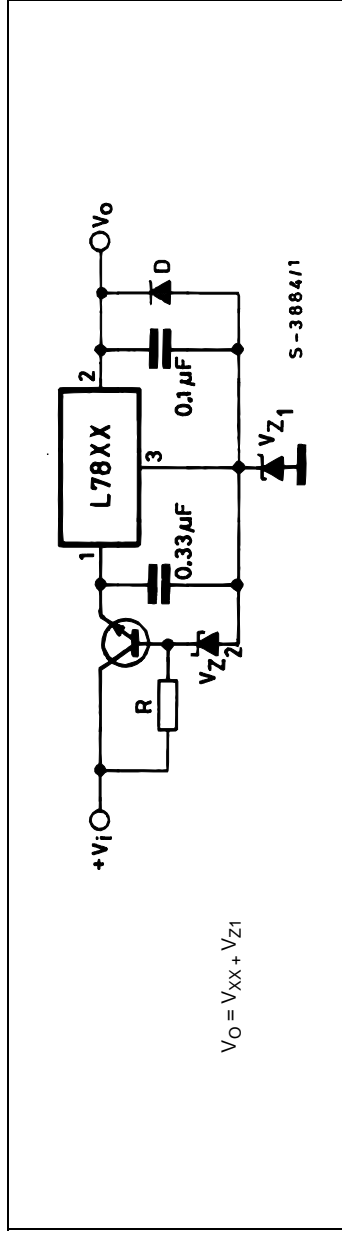
Figure 32: Reducing Power Dissipation with Dropping Resistor

$$R = \frac{V_{I(min)} - V_{XX} - V_{DROP(max)}}{I_{O(max)} + I_{d(max)}}$$
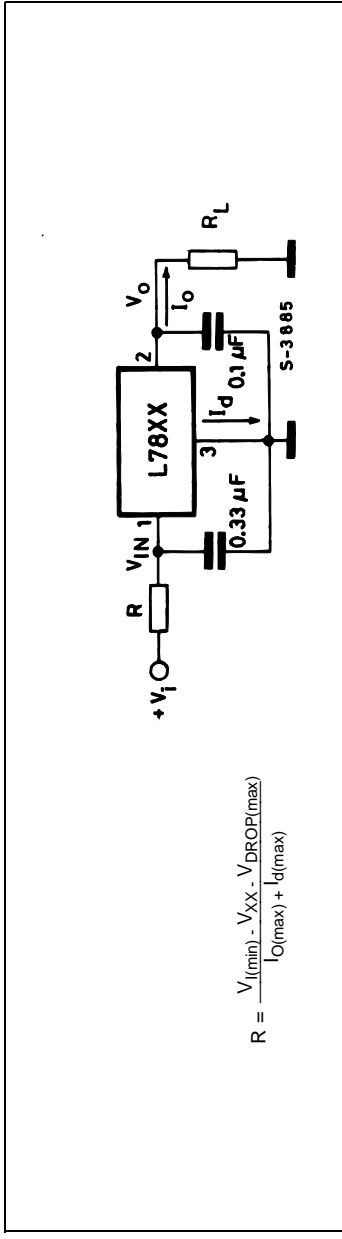
Figure 33: Remote Shutdown
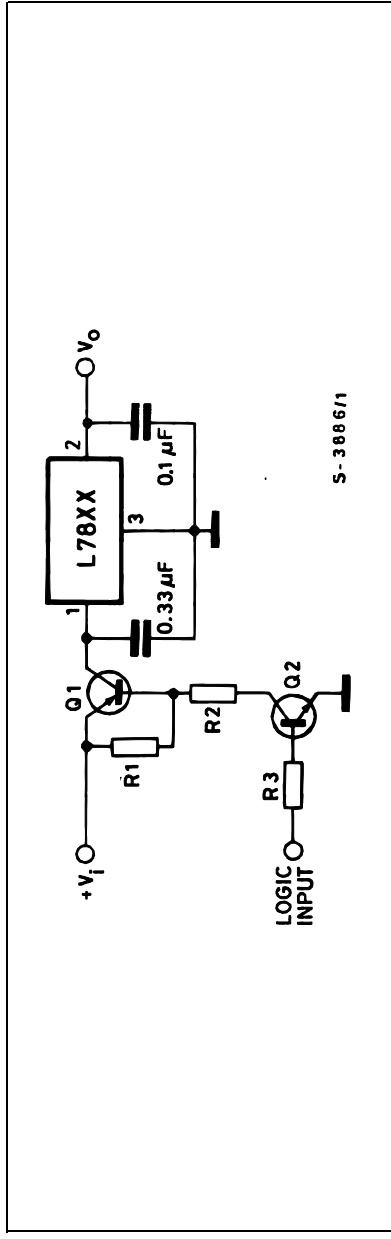
Figure 34: Power AM Modulator (unity voltage gain, $I_O \leq 0.5$)
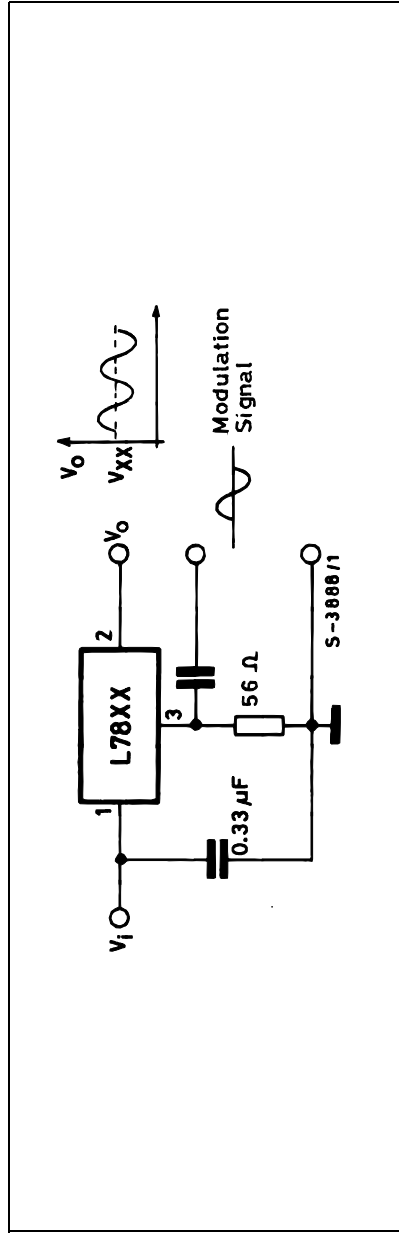
NOTE: The circuit performs well up to 100 KHz.

Figure 35: Adjustable Output Voltage with Temperature Compensation

$$V_O = V_{XX} \left(1 + \frac{R_2}{R_1}\right) + V_{BE}$$
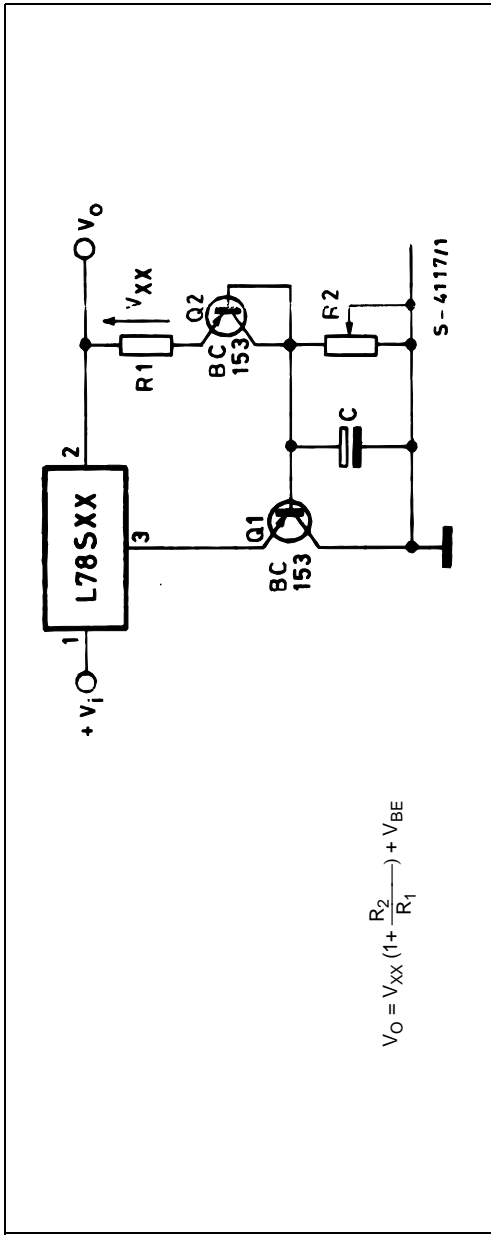
NOTE: Q_2 is connected as a diode in order to compensate the variation of the Q_1 V_{BE} with the temperature. C allows a slow rise time of the V_O.

Figure 36: Light Controllers ($V_{Omin} = V_{XX} + V_{BE}$)

V_O rises when the light goes up

V_O falls when the light goes up
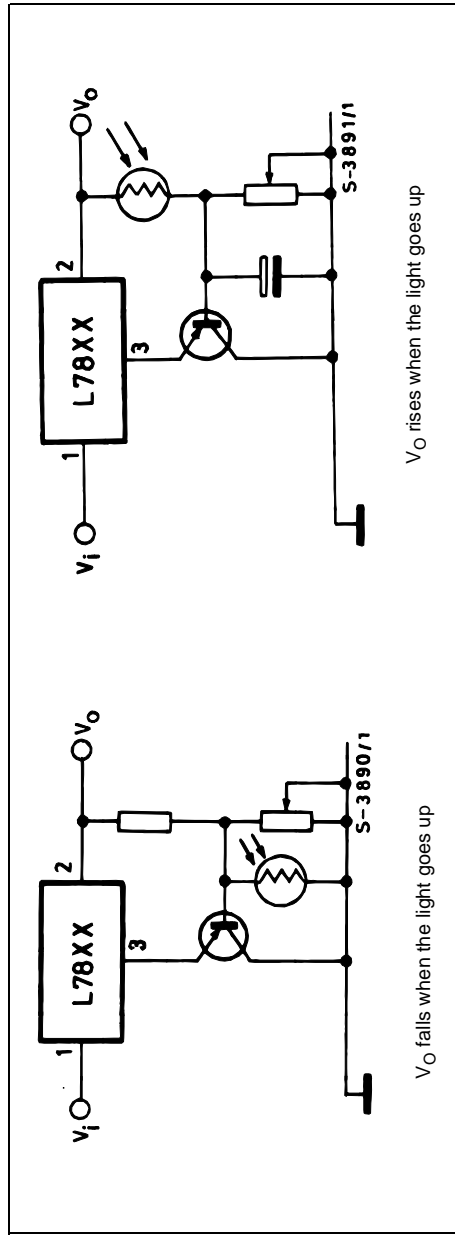
Figure 37: Protection against Input Short-Circuit with High Capacitance Loads
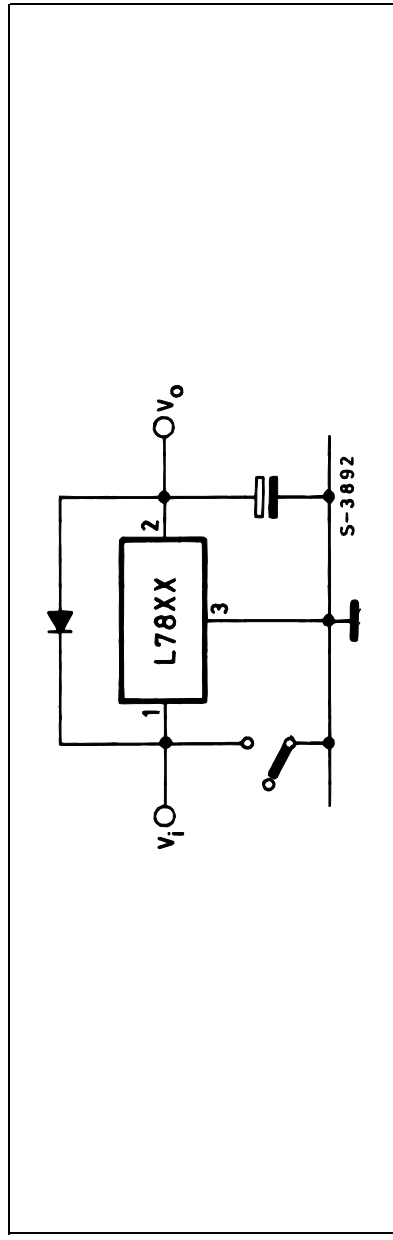
Application with high capacitance loads and an output voltage greater than 6 volts need an external diode (see fig. 33) to protect the device against input short circuit. In this case the input voltage falls rapidly while the output voltage decrease slowly. The capacitance discharges by means of the Base-Emitter junction of the series pass transistor in the regulator. If the energy is sufficiently high, the transistor may be destroyed. The external diode by-passes the current from the IC to ground.

TO-3 MECHANICAL DATA

DIM.	mm. MIN.	mm. TYP	mm. MAX.	inch MIN.	inch TYP.	inch MAX.
A		11.85			0.466	
B	0.96	1.05	1.10	0.037	0.041	0.043
C			1.70			0.066
D			8.7			0.342
E			20.0			0.787
G		10.9			0.429	
N		16.9			0.665	
P			26.2			1.031
R	3.88		4.09	0.152		0.161
U			39.5			1.555
V		30.10			1.185	

P003C/C

TO-220 (A TYPE) MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.40		4.60	0.173		0.181
b	0.61		0.88	0.024		0.034
b1	1.15		1.70	0.045		0.067
c	0.49		0.70	0.019		0.027
D	15.25		15.75	0.600		0.620
E	10.0		10.40	0.393		0.409
e	2.4		2.7	0.094		0.106
e1	4.95		5.15	0.194		0.203
F	1.23		1.32	0.048		0.051
H1	6.2		6.6	0.244		0.260
J1	2.40		2.72	0.094		0.107
L	13.0		14.0	0.511		0.551
L1	3.5		3.93	0.137		0.154
L20		16.4			0.645	
L30		28.9			1.138	
øP	3.75		3.85	0.147		0.151
Q	2.65		2.95	0.104		0.116

0015988/N

TO-220 (C TYPE) MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.30		4.70	0.169		0.185
b	0.70		0.90	0.028		0.035
b1	1.42		1.62	0.056		0.064
c	0.45		0.60	0.018		0.024
D		15.70			0.618	
E	9.80		10.20	0.386		0.402
e		2.54			0.100	
e1		5.08			0.200	
F	1.25		1.39	0.049		0.055
H1		6.5			0.256	
J1	2.20		2.60	0.087		0.202
L	12.88		13.28	0.507		0.523
L1		3			0.118	
L20	15.70		16.1	0.618		0.634
L30		28.9			1.138	
φP	3.50		3.70	0.138		0.146
Q	2.70		2.90	0.106		0.114
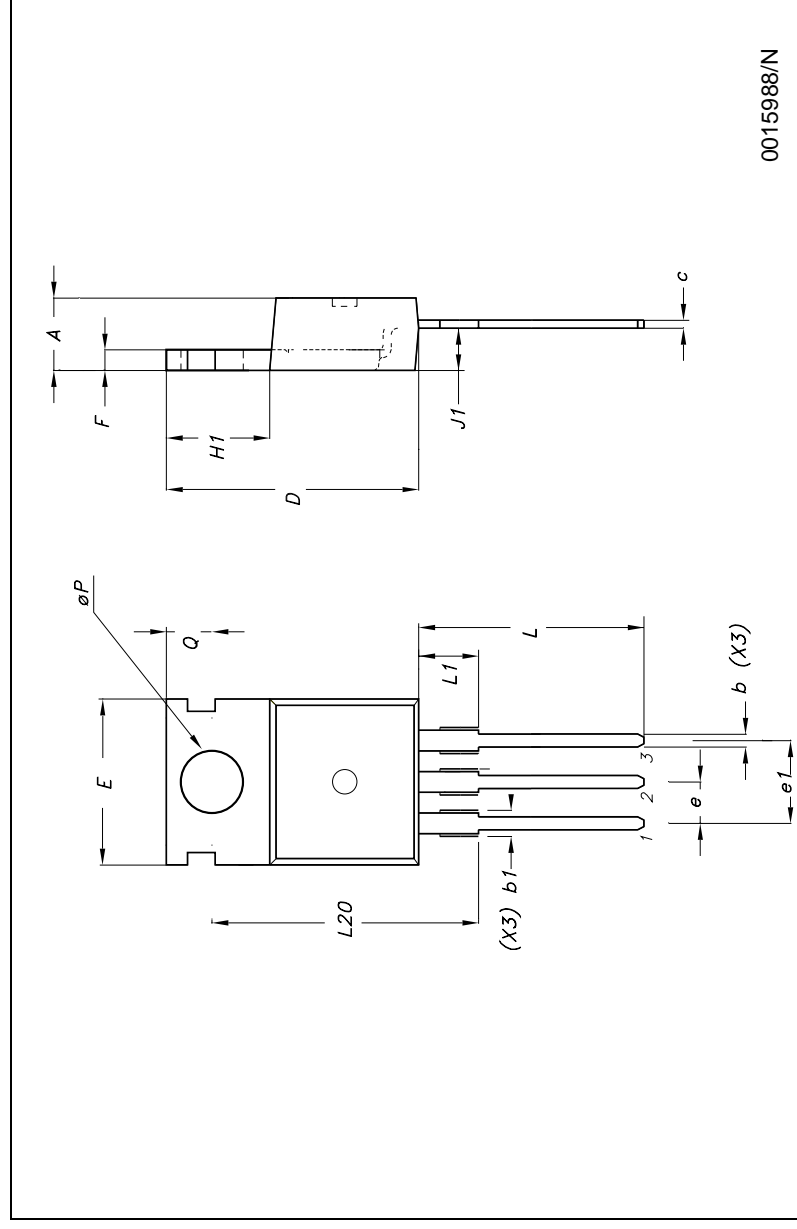
0015988/N

TO-220 (E TYPE) MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.47		4.67	0.176		0.184
b	0.70		0.91	0.028		0.036
b1	1.17		1.37	0.046		0.054
c	0.31		0.53	0.012		0.021
D	14.60		15.70	0.575		0.618
E	9.96		10.36	0.392		0.408
e		2.54			0.100	
e1		5.08			0.200	
F	1.17		1.37	0.046		0.054
H1	6.1		6.8	0.240		0.268
J1	2.52		2.82	0.099		0.111
L	12.70		13.80	0.500		0.543
L1	3.20		3.96	0.126		0.156
L20	15.21		16.77	0.599		0.660
øP	3.73		3.94	0.147		0.155
Q	2.59		2.89	0.102		0.114
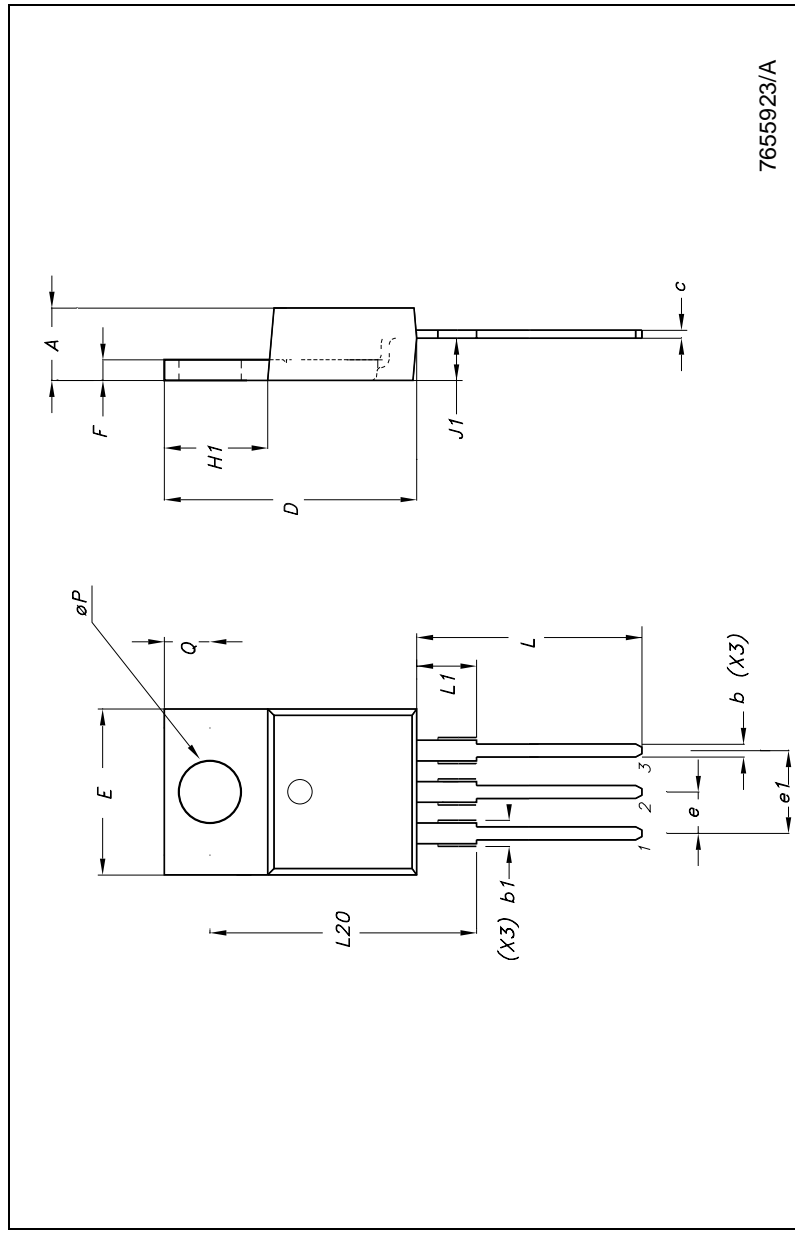
7655923/A

TO-220FP MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.40		4.60	0.173		0.181
B	2.5		2.7	0.098		0.106
D	2.5		2.75	0.098		0.108
E	0.45		0.70	0.017		0.027
F	0.75		1	0.030		0.039
F1	1.15		1.50	0.045		0.059
F2	1.15		1.50	0.045		0.059
G	4.95		5.2	0.194		0.204
G1	2.4		2.7	0.094		0.106
H	10.0		10.40	0.393		0.409
L2		16			0.630	
L3	28.6		30.6	1.126		1.204
L4	9.8		10.6	0.385		0.417
L5	2.9		3.6	0.114		0.142
L6	15.9		16.4	0.626		0.645
L7	9		9.3	0.354		0.366
DIA.	3		3.2	0.118		0.126
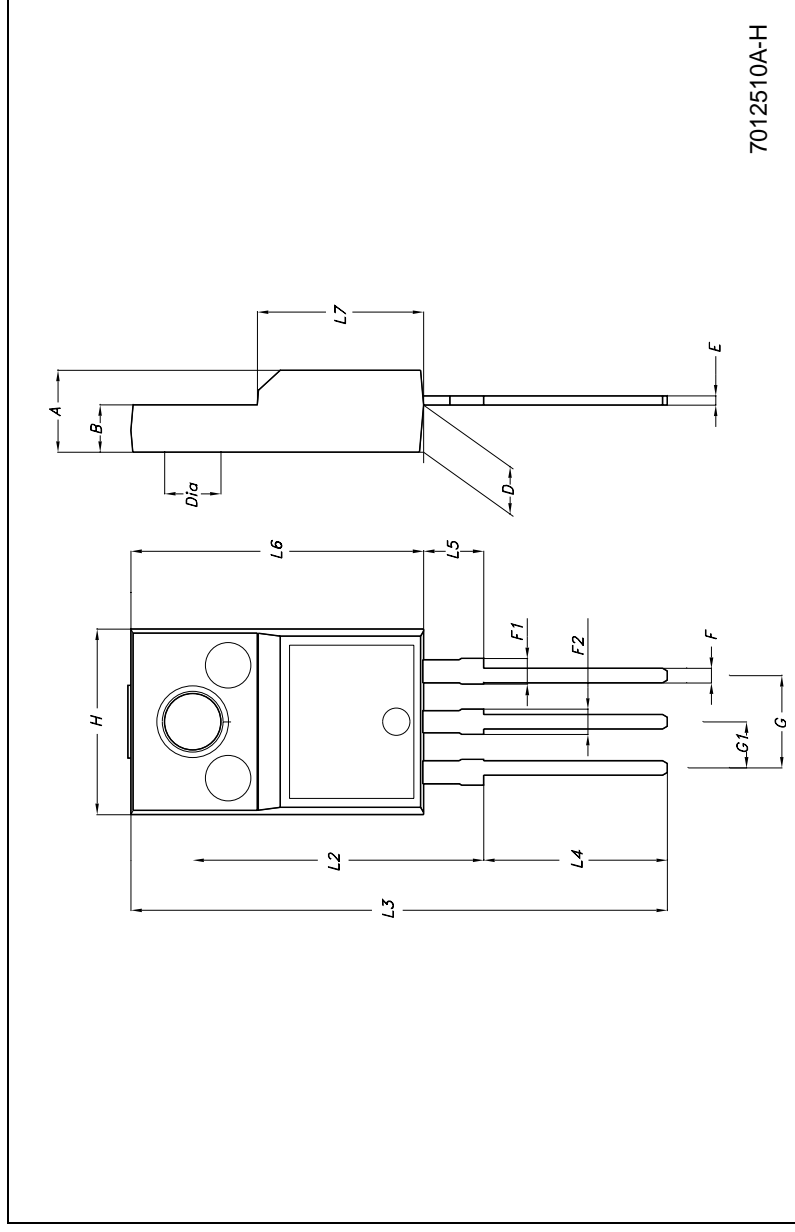
7012510A-H

TO-220FM MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.50		4.90	0.177		0.193
B	2.34		2.74	0.092		0.108
D	2.56		2.96	0.101		0.117
E	0.45	0.50	0.60	0.018	0.020	0.024
F	0.70		0.90	0.028		0.035
F1			1.47			0.058
G		5.08			0.200	
G1	2.34	2.54	2.74	0.092	0.100	0.108
H	9.96		10.36	0.392		0.408
L2		15.8			0.622	
L4	9.45		10.05	0.372		0.396
L6	15.67		16.07	0.617		0.633
L7	8.99		9.39	0.354		0.370
L8		3.30			0.130	
DIA.	3.08		3.28	0.121		0.129

7012510C-H
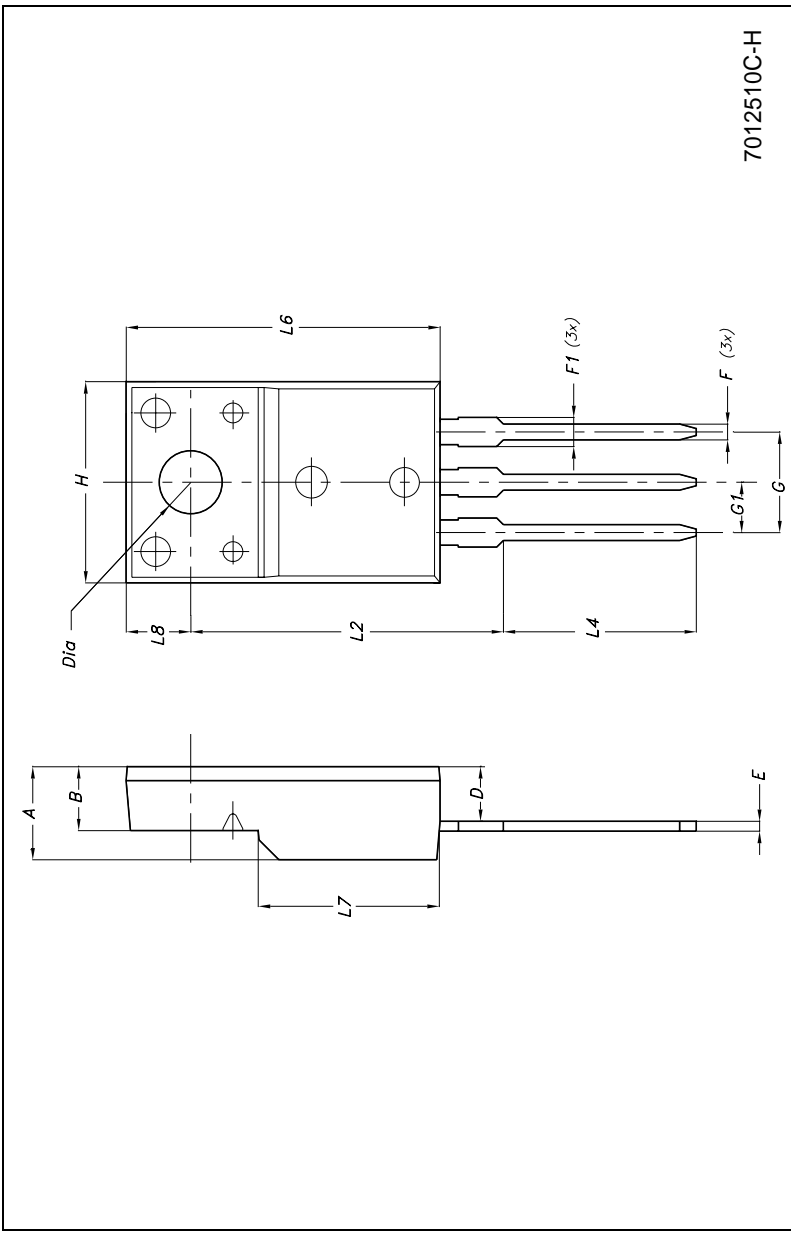
D²PAK (A TYPE) MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.4		4.6	0.173		0.181
A1	0.03		0.23	0.001		0.009
b	0.7		0.93	0.027		0.036
b2	1.14		1.7	0.044		0.067
c	0.45		0.6	0.017		0.023
c2	1.23		1.36	0.048		0.053
D	8.95		9.35	0.352		0.368
D1	8			0.315		
E	10		10.4	0.393		0.409
E1	8.5			0.335		
e		2.54			0.100	
e1	4.88		5.28	0.192		0.208
H	15		15.85	0.590		0.624
J1	2.49		2.69	0.098		0.106
L	2.29		2.79	0.090		0.110
L1	1.27		1.4	0.050		0.055
L2	1.3		1.75	0.051		0.069
R		0.4			0.016	
V2	0°		8°	0°		8°
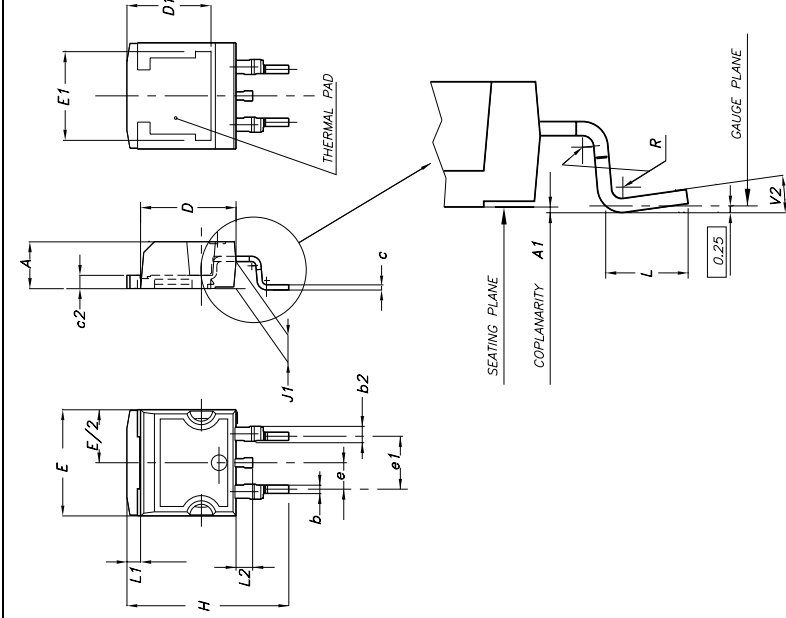
0079457/J

D²PAK (C TYPE) MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A	4.3		4.7	0.169		0.185
A1	0		0.20	0.000		0.008
b	0.70		0.90	0.028		0.035
b2	1.17		1.37	0.046		0.054
c	0.45	0.50	0.6	0.018	0.020	0.024
c2	1.25	1.30	1.40	0.049	0.051	0.055
D	9.0	9.2	9.4	0.354	0.362	0.370
D1	7.5			0.295		
E	9.8		10.2	0.386		0.402
E1	7.5			0.295		
e		2.54			0.100	
e1		5.08			0.200	
H	15	15.30	15.60	0.591	0.602	0.614
J1	2.20		2.60	0.087		0.102
L	1.79		2.79	0.070		0.110
L1	1.0		1.4	0.039		0.055
L2	1.2		1.6	0.047		0.063
R		0.3			0.012	
V2	0°		3°	0°		3°
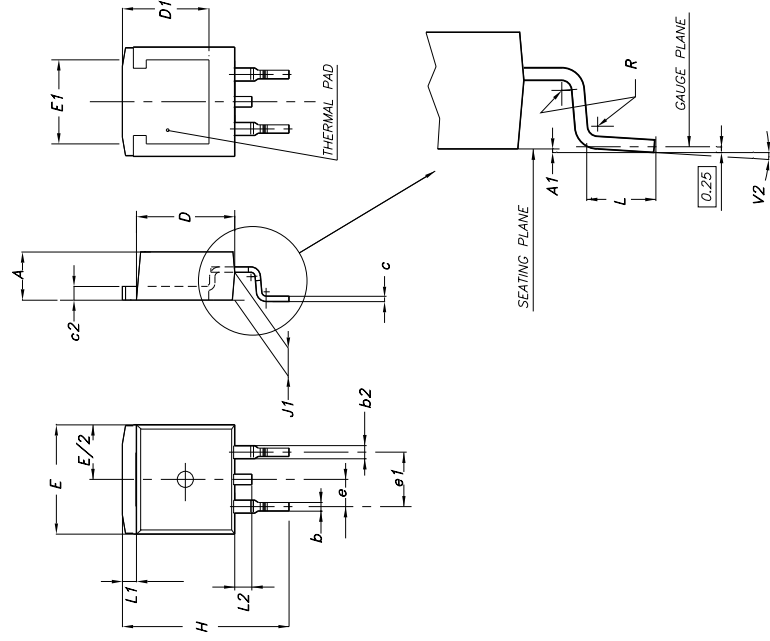
0079457/J

Tape & Reel D²PAK-P²PAK-D²PAK/A-P²PAK/A MECHANICAL DATA

DIM.	mm.			inch		
	MIN.	TYP	MAX.	MIN.	TYP.	MAX.
A			180			7.086
C	12.8	13.0	13.2	0.504	0.512	0.519
D	20.2			0.795		
N	60			2.362		
T			14.4			0.567
Ao	10.50	10.6	10.70	0.413	0.417	0.421
Bo	15.70	15.80	15.90	0.618	0.622	0.626
Ko	4.80	4.90	5.00	0.189	0.193	0.197
Po	3.9	4.0	4.1	0.153	0.157	0.161
P	11.9	12.0	12.1	0.468	0.472	0.476

Note: Drawing not in scale

Table 24: Revision History

Date	Revision	Description of Changes
09-Nov-2004	12	Add New Part Number.

Appendix E-10: DC Power Jack Specifications

RoHS

Ø1.5[0.06]

2.5[0.10] (2 PLCS)
3.0[0.12]

1
3
2

3.0[0.12]
10.7[0.42]

14.4[0.57]
3.50[0.14]

11.0[0.43]
6.50[0.26]
3.5[0.14]
4.7[0.19]

9.0[0.35]
Ø2.00[0.08]
Ø6.50[0.26]
3.00[0.12]

10.7[0.42]
3.0[0.12]

1.00X3.00 (2 PLCS)

2
3
1

3.0[0.12]
4.7[0.19]
1.00X3.50

PCB LAYOUT
TOP VIEW

| MODEL NO. | PJ-202A |

SCHEMATIC

1
3
2

CENTER PIN DIAMETER 2.0mm Dia.

PC FILE NAME:
PJ-202A

SPECIFICATIONS:
RATING: 16V DC @ 2.5A
CONTACT RESISTANCE: 50m OHMS MAX
INSULATION RESISTANCE: 50M OHMS MIN: 100V DC
VOLTAGE WITHSTAND: 500V AC R.M.S. FOR 1 MINUTE
LIFE: 5,000 CYCLES

	MATERIAL	PLATING
CENTER PIN	Copper	Nickel
TERMINAL 1	Brass	Tin
TERMINAL 2	Copper Alloy	Tin
TERMINAL 3	Brass	Tin
HOUSING	PBT	

TOLERANCE:
X.X ±0.2mm
X.XX ±0.1mm
X.XXX ±0.05mm

CUI INC
9615 SW Allen Blvd., Ste. 103
Beaverton, OR 97005
Phone: 503-643-4899
800-275-4899
Fax: 503-372-1266
Website: www.cui.com

TITLE: DC POWER JACK
PART NO. PJ-202A
DRAWN BY: ZRJ
APPROVED BY:
REV: A
UNITS: MM [INCHES]
SCALE: 2:1

Appendix E-11: LED Specifications

INNOVATIVE TECHNOLOGIES
WHERE INNOVATION COMES TO LIGHT

CMD15-21 & 22 Series SMT LEDs 1206 Package Size

DESCRIPTION AND FEATURES
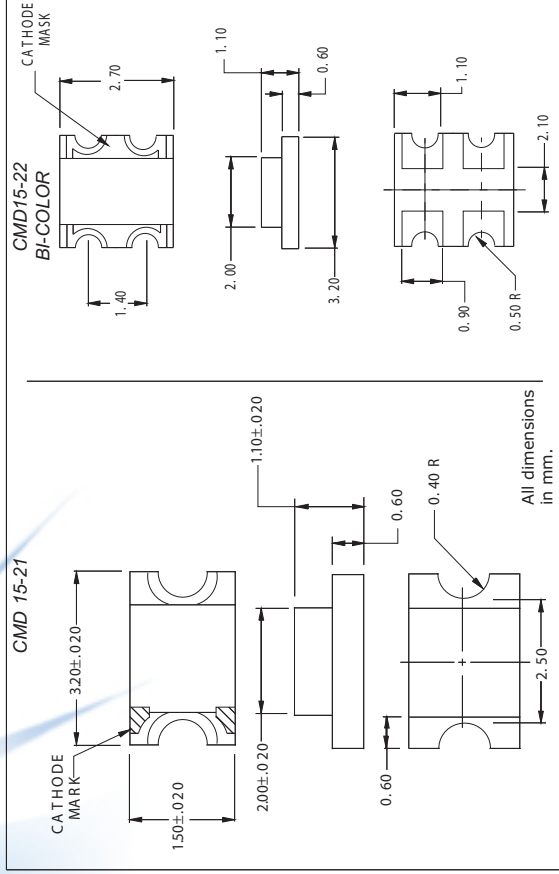
Surface Mount Technology
- Tape and reel packaged for high-speed autoinsertion.
- Convection and vapor-phase reflow compatible.
- Compact form enables high density placement.
- Packaged 2000 pieces per reel.

Leading Edge LED Optoelectronic Performance
- Consistent high brightness.
- Integral lensing for narrow angle light dispersion.
- Bi-color available.

Exceptional Reliability
- Stringent process controls assure quality.
- Extensive qualification testing to meet strictest requirements.
- Designed to permit easy post-reflow solder joint inspection.
- Solder pad geometry on page 1-X2.

CMD 15-21

CATHODE MARK

320±.020
1.10±.020
0.60
0.40 R
2.50
2.00±.020
1.50±.020
0.60

CMD15-22 BI-COLOR

CATHODE MASK

2.70
1.40
1.10
0.60
2.00
3.20
1.10
2.10
0.90
0.50 R

All dimensions in mm.

ELECTRO-OPTICAL CHARACTERISTICS

Part Number	Emitted Color	Lens Color	Luminous Intensity Min. (mcd)	Luminous Intensity Typ. (mcd)	Forward Voltage Typ. (V)	Forward Voltage Max. (V)	Test Current (mA)	Peak Wavelength (nm)
CMD15-21S RC/TR8	Super Red	Clear	6.0	15.0	1.7	2.4	20	660
CMD15-21V RC/TR8	H.E.Red	Clear	3.6	6.0	2.0	2.8	20	640
CMD15-21V YC/TR8	Yellow	Clear	3.6	6.0	2.0	2.8	20	585
CMD15-21V GC/TR8	Green	Clear	6.0	10.0	2.1	2.8	20	570
CMD15-21U BC/TR8	Blue	Clear	19.0	45.0	4.9	5.5	20	430
CMD15-21U GC/TR8	Green	Clear	7.0	12.0	2.1	2.8	20	565
CMD15-21V RD/TR8	H.E.Red	Red Diffused	2.4	4.0	2.0	2.8	20	640
CMD15-21V GD/TR8	Clear	Green Diffused	5.5	6.5	2.1	2.8	20	570
CMD15-21V YD/TR8	Clear	Yellow Diffused	3.7	4.0	2.0	2.8	20	585
CMD15-21U YC/TR8	Super Yellow	Clear	27.0	47.0	2.1	2.8	20	590
CMD1521U SOC/TR8	Sunset orange	Clear	17.0	31.0	2.1	2.8	20	620
CMD1522V RVGC/TR8	H.E.Red	Clear	3.6	20.0	2.8	2.8	20	640
	Green	Clear	6.0	10.0	2.1	2.8	20	570
CMD15-22S RUGC/TR8	Red	Clear	9.0	15.0	1.7	2.4	20	660
	Green	Clear	7.0	12.0	2.1	2.8	20	565

ABSOLUTE MAXIMUM RATINGS

	Green	H.E.Red	Yellow	Red	Units
Power Dissipation	105	105	105	100	mW
Storage/Operating Temperature	-40 to +85	-40 to +85	-40 to +85	-40 to +85	°C
Peak Forward Current (1µs @ 10% duty cycle)	150	150	150	150	mA
Reverse Voltage (IR=100µA)	5.0	5.0	5.0	5.0	V
Lead Solder Time @ 260°C	5	5	5	5	Seconds
Average Forward Current	25	30	30	30	mA

INNOVATIVE TECHNOLOGIES
WHERE INNOVATION COMES TO LIGHT

CML-IT reserves the right to make specification revisions that enhance the design and/or performance of the product

Appendix E-12: Transistor Specifications

2N3904

TO-92

C B E

MMBT3904

C

E

B
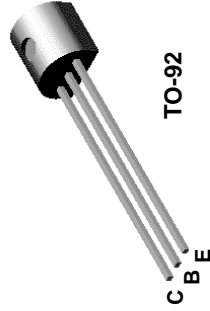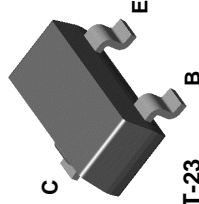
SOT-23
Mark: 1A

PZT3904

C

E

C

B
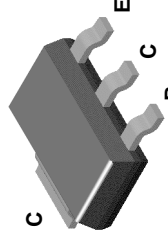
SOT-223

NPN General Purpose Amplifier

This device is designed as a general purpose amplifier and switch.
The useful dynamic range extends to 100 mA as a switch and to
100 MHz as an amplifier.

Absolute Maximum Ratings* T_A = 25°C unless otherwise noted

Symbol	Parameter	Value	Units
V_{CEO}	Collector-Emitter Voltage	40	V
V_{CBO}	Collector-Base Voltage	60	V
V_{EBO}	Emitter-Base Voltage	6.0	V
I_C	Collector Current - Continuous	200	mA
T_J, T_{stg}	Operating and Storage Junction Temperature Range	-55 to +150	°C

*These ratings are limiting values above which the serviceability of any semiconductor device may be impaired.

NOTES:
1) These ratings are based on a maximum junction temperature of 150 degrees C.
2) These are steady state limits. The factory should be consulted on applications involving pulsed or low duty cycle operations.

Thermal Characteristics T_A = 25°C unless otherwise noted

Symbol	Characteristic		Max		Units
		2N3904	***MMBT3904**	****PZT3904**	
P_D	Total Device Dissipation	625	350	1,000	mW
	Derate above 25°C	5.0	2.8	8.0	mW/°C
$R_{\theta JC}$	Thermal Resistance, Junction to Case	83.3			°C/W
$R_{\theta JA}$	Thermal Resistance, Junction to Ambient	200	357	125	°C/W

*Device mounted on FR-4 PCB 1.6" X 1.6" X 0.06."
**Device mounted on FR-4 PCB 36 mm X 18 mm X 1.5 mm; mounting pad for the collector lead min. 6 cm².
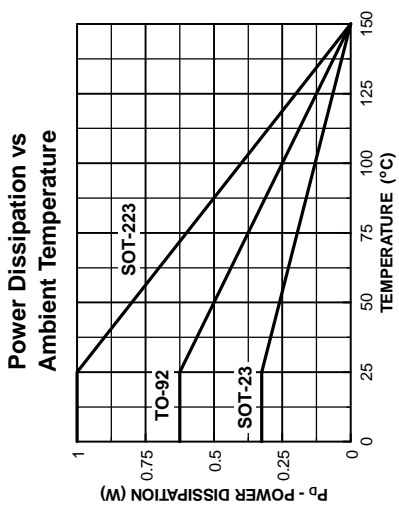
2N3904 / MMBT3904 / PZT3904

NPN General Purpose Amplifier
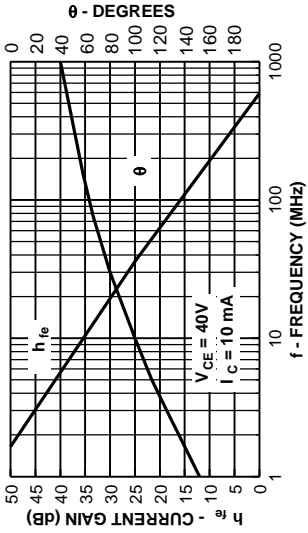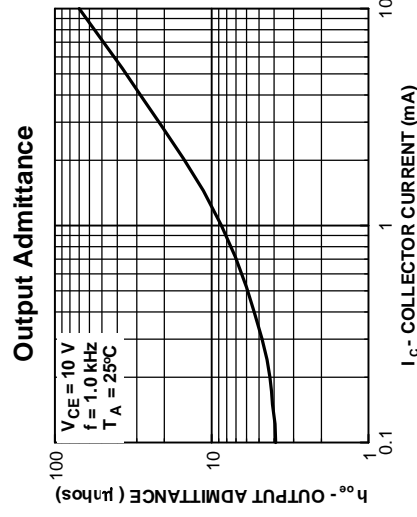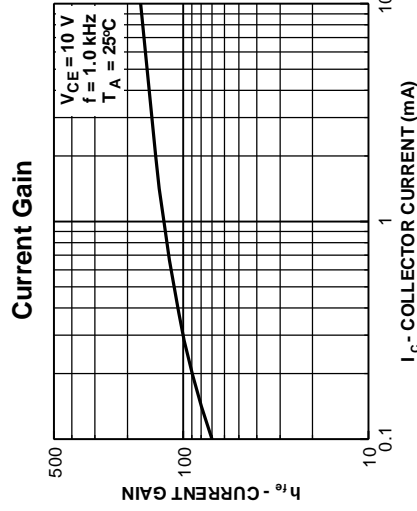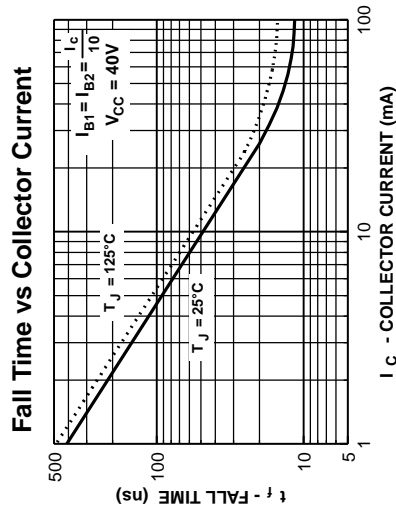(continued)

Electrical Characteristics T_A = 25°C unless otherwise noted

Symbol	Parameter	Test Conditions	Min	Max	Units
OFF CHARACTERISTICS					
$V_{(BR)CEO}$	Collector-Emitter Breakdown Voltage	I_C = 1.0 mA, I_B = 0	40		V
$V_{(BR)CBO}$	Collector-Base Breakdown Voltage	I_C = 10 μA, I_E = 0	60		V
$V_{(BR)EBO}$	Emitter-Base Breakdown Voltage	I_E = 10 μA, I_C = 0	6.0		V
I_{BL}	Base Cutoff Current	V_{CE} = 30 V, V_{EB} = 3V		50	nA
I_{CEX}	Collector Cutoff Current	V_{CE} = 30 V, V_{EB} = 3V		50	nA
ON CHARACTERISTICS*					
h_{FE}	DC Current Gain	I_C = 0.1 mA, V_{CE} = 1.0 V	40		
		I_C = 1.0 mA, V_{CE} = 1.0 V	70		
		I_C = 10 mA, V_{CE} = 1.0 V	100	300	
		I_C = 50 mA, V_{CE} = 1.0 V	60		
		I_C = 100 mA, V_{CE} = 1.0 V	30		
$V_{CE(sat)}$	Collector-Emitter Saturation Voltage	I_C = 10 mA, I_B = 1.0 mA		0.2	V
		I_C = 50 mA, I_B = 5.0 mA		0.3	V
$V_{BE(sat)}$	Base-Emitter Saturation Voltage	I_C = 10 mA, I_B = 1.0 mA	0.65	0.85	V
		I_C = 50 mA, I_B = 5.0 mA		0.95	V
SMALL SIGNAL CHARACTERISTICS					
f_T	Current Gain - Bandwidth Product	I_C = 10 mA, V_{CE} = 20 V, f = 100 MHz	300		MHz
C_{obo}	Output Capacitance	V_{CB} = 5.0 V, I_E = 0, f = 1.0 MHz		4.0	pF
C_{ibo}	Input Capacitance	V_{EB} = 0.5 V, I_C = 0, f = 1.0 MHz		8.0	pF
NF	Noise Figure	I_C = 100 μA, V_{CE} = 5.0 V, R_S =1.0kΩ,f=10 Hz to 15.7kHz		5.0	dB
SWITCHING CHARACTERISTICS					
t_d	Delay Time	V_{CC} = 3.0 V, V_{BE} = 0.5 V,		35	ns
t_r	Rise Time	I_C = 10 mA, I_{B1} = 1.0 mA		35	ns
t_s	Storage Time	V_{CC} = 3.0 V, I_C = 10mA		200	ns
t_f	Fall Time	I_{B1} = I_{B2} = 1.0 mA		50	ns

*Pulse Test: Pulse Width ≤ 300 μs, Duty Cycle ≤ 2.0%

Spice Model

NPN (Is=6.734f Xti=3 Eg=1.11 Vaf=74.03 Bf=416.4 Ne=1.259 Ise=6.734 Ikf=66.78m Xtb=1.5 Br=.7371 Nc=2 Isc=0 Ikr=0 Rc=1 Cjc=3.638p Mjc=.3085 Vjc=.75 Fc=.5 Cje=4.493p Mje=.2593 Vje=.75 Tr=239.5n Tf=301.2p Itf=.4 Vtf=4 Xtf=2 Rb=10)

2N3904 / MMBT3904 / PZT3904

NPN General Purpose Amplifier
(continued)

Typical Characteristics

Collector-Emitter Saturation Voltage vs Collector Current

Base-Emitter ON Voltage vs Collector Current

Capacitance vs Reverse Bias Voltage

Typical Pulsed Current Gain vs Collector Current

Base-Emitter Saturation Voltage vs Collector Current

Collector-Cutoff Current vs Ambient Temperature

NPN General Purpose Amplifier (continued)

Typical Characteristics (continued)

Noise Figure vs Source Resistance

NF - NOISE FIGURE (dB)

R_S - SOURCE RESISTANCE (kΩ)

$I_C = 50\ \mu A$
$I_C = 100\ \mu A$
$I_C = 1.0\ mA$
$I_C = 5.0\ mA$

Power Dissipation vs Ambient Temperature

P_D - POWER DISSIPATION (W)

TEMPERATURE (°C)

SOT-223
TO-92
SOT-23

Rise Time vs Collector Current

t_r - RISE TIME (ns)

I_C - COLLECTOR CURRENT (mA)

$I_C / I_{B1} = I_{B2} = 10$
$V_{CC} = 40V$
$T_J = 25°C$
$T_J = 125°C$

Noise Figure vs Frequency

NF - NOISE FIGURE (dB)

f - FREQUENCY (kHz)

$V_{CE} = 5.0V$
$I_C = 1.0\ mA$ $R_S = 200\Omega$
$I_C = 50\ \mu A$ $R_S = 1.0\ k\Omega$
$I_C = 0.5\ mA$ $R_S = 200\Omega$
$I_C = 100\ \mu A, R_S = 500\ \Omega$

Current Gain and Phase Angle vs Frequency

θ - DEGREES

h_{fe} - CURRENT GAIN (dB)

f - FREQUENCY (MHz)

h_{fe}
θ
$V_{CE} = 40V$
$I_C = 10\ mA$

Turn-On Time vs Collector Current

TIME (ns)

I_C - COLLECTOR CURRENT (mA)

$I_C / I_{B1} = I_{B2} = 10$
$t_r @ V_{CC} = 3.0V$
$t_d @ V_{CB} = 0V$
15V
40V
2.0V

2N3904 / MMBT3904 / PZT3904

NPN General Purpose Amplifier (continued)

Typical Characteristics (continued)

Fall Time vs Collector Current

$I_{B1} = I_{B2} = \dfrac{I_C}{10}$

$V_{CC} = 40V$

$T_J = 125°C$

$T_J = 25°C$

t_f - FALL TIME (ns)

I_C - COLLECTOR CURRENT (mA)

Output Admittance

$V_{CE} = 10\ V$
$f = 1.0\ kHz$
$T_A = 25°C$

h_{oe} - OUTPUT ADMITTANCE ($\mu mhos$)

I_C - COLLECTOR CURRENT (mA)

Voltage Feedback Ratio

$V_{CE} = 10\ V$
$f = 1.0\ kHz$
$T_A = 25°C$

h_{re} - VOLTAGE FEEDBACK RATIO (x10^{-4})

I_C - COLLECTOR CURRENT (mA)

Storage Time vs Collector Current

$I_{B1} = I_{B2} = \dfrac{I_C}{10}$

$T_J = 25°C$

$T_J = 125°C$

t_s - STORAGE TIME (ns)

I_C - COLLECTOR CURRENT (mA)

Current Gain

$V_{CE} = 10\ V$
$f = 1.0\ kHz$
$T_A = 25°C$

h_{fe} - CURRENT GAIN

I_C - COLLECTOR CURRENT (mA)

Input Impedance

$V_{CE} = 10\ V$
$f = 1.0\ kHz$
$T_A = 25°C$

h_{ie} - INPUT IMPEDANCE (kΩ)

I_C - COLLECTOR CURRENT (mA)

2N3904 / MMBT3904 / PZT3904

NPN General Purpose Amplifier
(continued)

Test Circuits
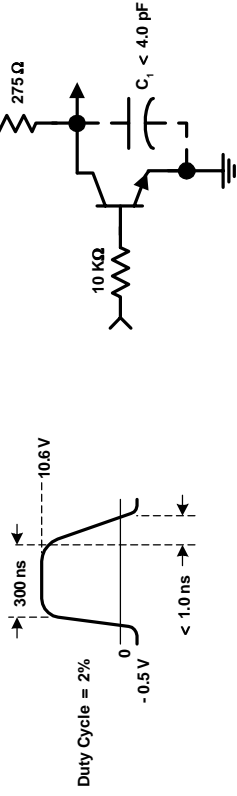
FIGURE 1: Delay and Rise Time Equivalent Test Circuit
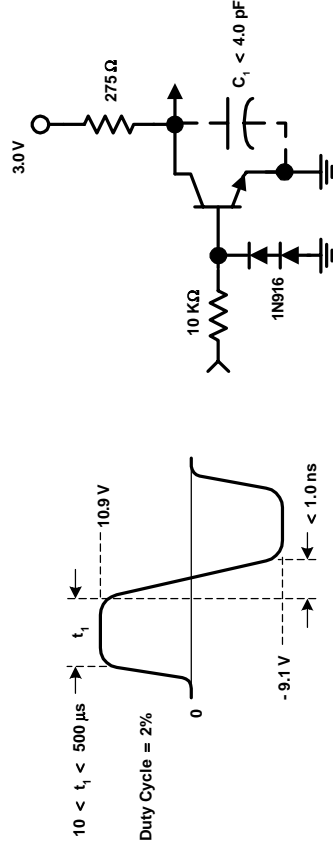
FIGURE 2: Storage and Fall Time Equivalent Test Circuit

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ACEx™	FASTr™	PowerTrench®
Bottomless™	GlobalOptoisolator™	QFET™
CoolFET™	GTO™	QS™
CROSSVOLT™	HiSeC™	QT Optoelectronics™
DOME™	ISOPLANAR™	Quiet Series™
E²CMOS™	MICROWIRE™	SILENT SWITCHER®
EnSigna™	OPTOLOGIC™	SMART START™
FACT™	OPTOPLANAR™	SuperSOT™-3
FACT Quiet Series™	PACMAN™	SuperSOT™-6
FAST®	POP™	SuperSOT™-8

SyncFET™
TinyLogic™
UHC™
VCX™

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2. A critical component is any component of a life support device or system whose failure to perform can be reasonably expected to cause the failure of the life support device or system, or to affect its safety or effectiveness.

PRODUCT STATUS DEFINITIONS
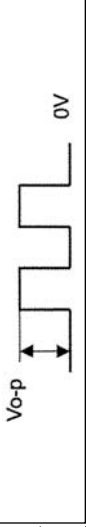
Definition of Terms

Datasheet Identification	Product Status	Definition
Advance Information	Formative or In Design	This datasheet contains the design specifications for product development. Specifications may change in any manner without notice.
Preliminary	First Production	This datasheet contains preliminary data, and supplementary data will be published at a later date. Fairchild Semiconductor reserves the right to make changes at any time without notice in order to improve design.
No Identification Needed	Full Production	This datasheet contains final specifications. Fairchild Semiconductor reserves the right to make changes at any time without notice in order to improve design.
Obsolete	Not In Production	This datasheet contains specifications on a product that has been discontinued by Fairchild semiconductor. The datasheet is printed for reference information only.

Appendix E-13: Buzzer Specifications

CUI INC

RoHS

Specifications

Rated voltage	3.5 Vo-p
Operating voltage	3.0 - 5.0 Vo-p
Mean current	35 mA max.	Applying rated voltage, 2048 Hz square wave, ½ duty
Coil resistance	42 ±6.3 Ω
Sound output	Min. 85 (Typical 95) dBA	Distance at 10cm (A-weight free air). Applying rated voltage of 2048 Hz, square wave, 1/2 duty.
Rated frequency	2,048 Hz
Operating temperature	-20 ~ +60° C
Storage temperature	-30 ~ +70° C
Dimensions	⌀12.0 x H8.5 mm	See attached drawing
Weight	1.4 g
Material	PPO (Black)
Terminal	Pin type (AU Plating)	See attached drawing
RoHS	yes

Vo-p

0v

Frequency Response Curve

A: Frequency Response. Magn dB re 20.00uPA

CUI INC

Part No: CEM-1203(42)

Description: magnetic buzzer

Date: 8/11/2006

Unit: mm

Page No: 2 of 5
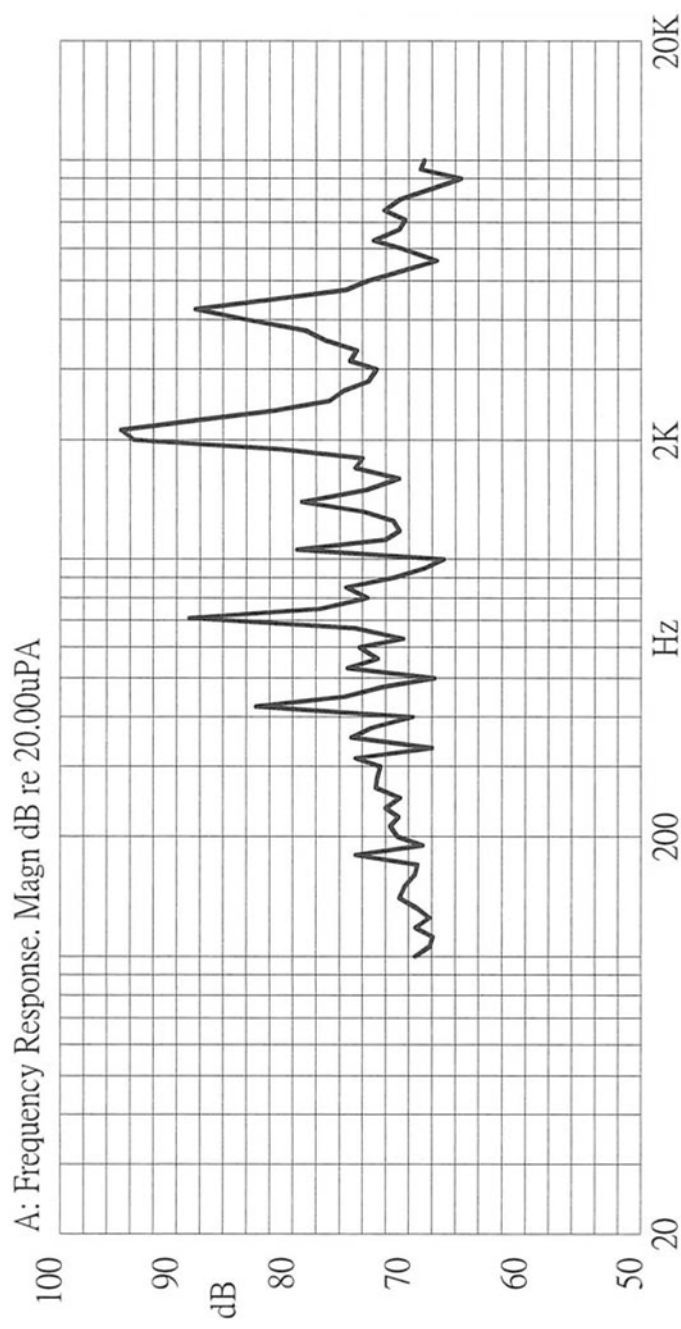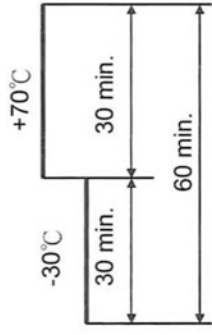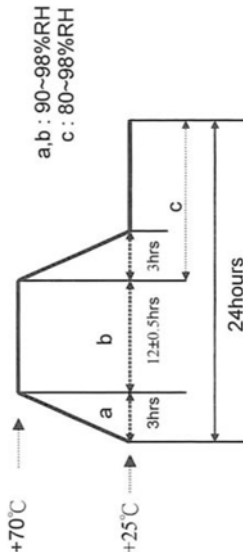
Appearance Drawing

Tolerance: ±0.5

6.5

2-ø0.6±0.1

5.5

8.5

5.5

(−)   (+)

Sound
Emission
Hole ø2.3

CEM-1203(42) CUI INC

ø12.0

Measurement Method

+VDC

TR:
2SC1741AS

180Ω

+2.5V

0

CUI INC

Part No: CEM-1203(42)

Description: magnetic buzzer

Date: 8/11/2006
Unit: mm
Page No: 3 of 5
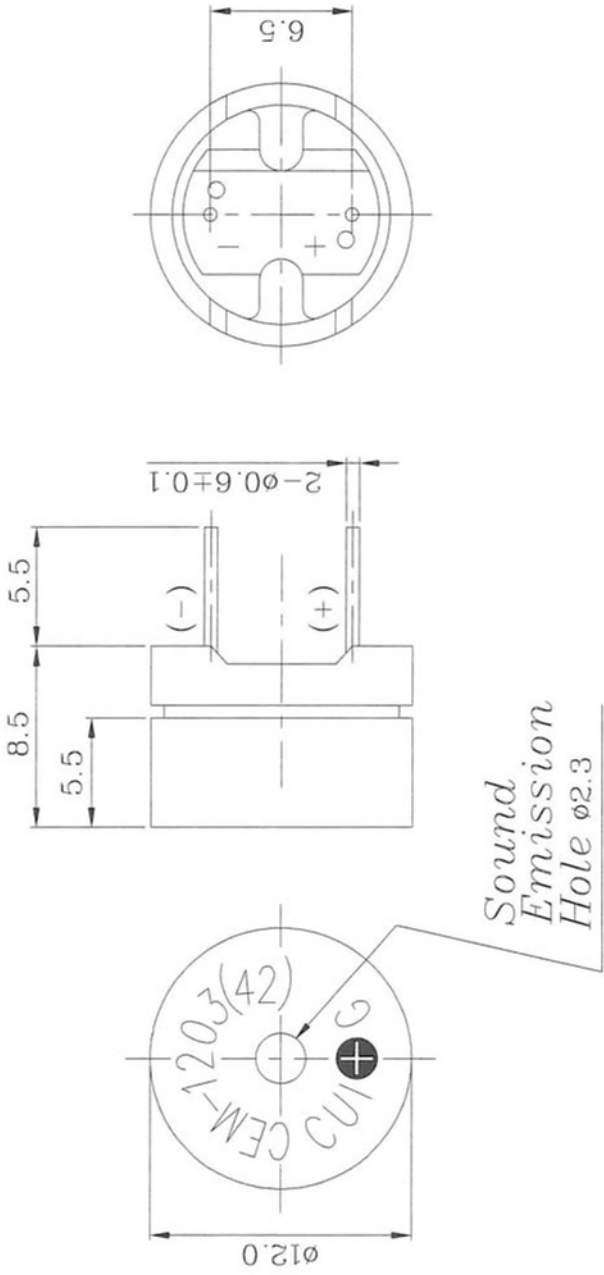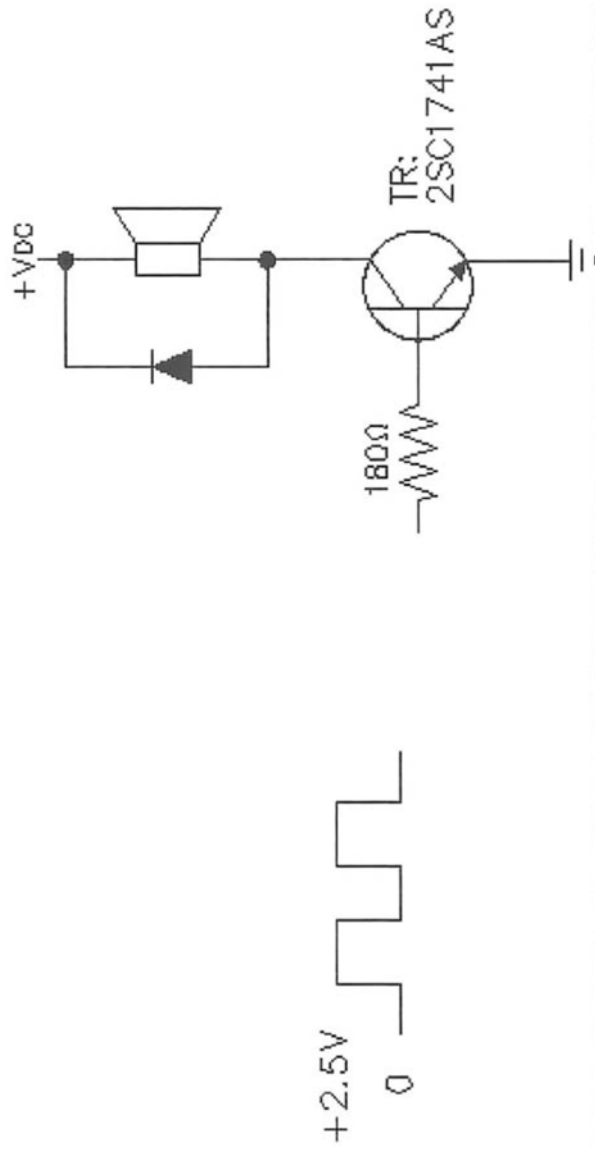
Mechanical Characteristics

Item	Test Condition	Evaluation Standard
Solderability	Lead terminals are immersed in rosin for 5 seconds and then immersed in solder bath of 270 ±5°C for 3 ±1 seconds.	90% min. of lead terminals should be wet with solder. (Except the edge of the terminal)
Soldering Heat Resistance	Lead terminals are immersed up to 1.5mm from the buzzer's body in solder bath of 260 ±5°C for 3 ±1 seconds.	No in interference in operation.
Terminal Mechanical Strength	Apply force of 9.8 N (1.0 kg) to each terminal in each axial direction for 10 seconds.	No damage or cutting off.
Vibration	The buzzer will be measured after applying a vibration amplitude of 1.5 mm with 10 to 55 Hz band of vibration frequency to each of the 3 perpendicular directions for 2 hours.	After the test, the part shall meet specifications without any damage to the appearance. After 4 hours at +25°C, the SPL should be within ±10 dBA of the initial SPL.
Drop Test	The part is to be dropped from a height of 75 cm onto a 40 mm thick wooden board 3 times in 3 axis (X, Y, Z) for a total of 9 drops.	

Environment Test

Item	Test Condition	Evaluation Standard
High temp. test	The part will be subjected to +70°C for 96 hours.	
Low temp. test	The part will be subjected to -30°C for 96 hours.	
Thermal shock	The part will be subjected to 10 cycles. One cycle will consist of:	
Temp./Humidity cycle	The part shall be subjected to 10 cycles. One cycle will consist of:	After the test, the part shall meet specifications without any damage to the appearance. After 4 hours at +25°C, the SPL should be within ±10 dBA of the initial SPL.

CUI INC

Reliability Tests

Item	Test Condition	Evaluation Standard
Operating (Life Test)	1. Continous life test: The part will be subjected to 72 hours at 45°C with 3.5 V, 2048 Hz applied. 2. Intermittent life test: A duty cycle of 1 minute on, 1 mintue off, a minimum of 10,000 times at room temp. (25±10°C) with 3.5 V, 2048 Hz applied.	After the test, the part shall meet specifications without any damage to the appearance. After 4 hours at +25°C, the SPL should be within ±10 dBA of the initial SPL.

Test Conditions

Standard Test Condition	a) Tempurature: +5 ~ +35°C	b) Humidity: 45 - 85%	c) Pressure: 860 - 1060 mbar
Judgement Test Condition	a) Tempurature: +25±2°C	b) Humidity: 60 - 70%	c) Pressure: 860 - 1060 mbar

CUI INC

Packaging

100PCS

TRAY

Out Box
(5PCS Tray)

Carton Box
(12PCS Out Box)

310mm

440mm

400mm

1. CEM-1203(42)

2. RoHS Compliant
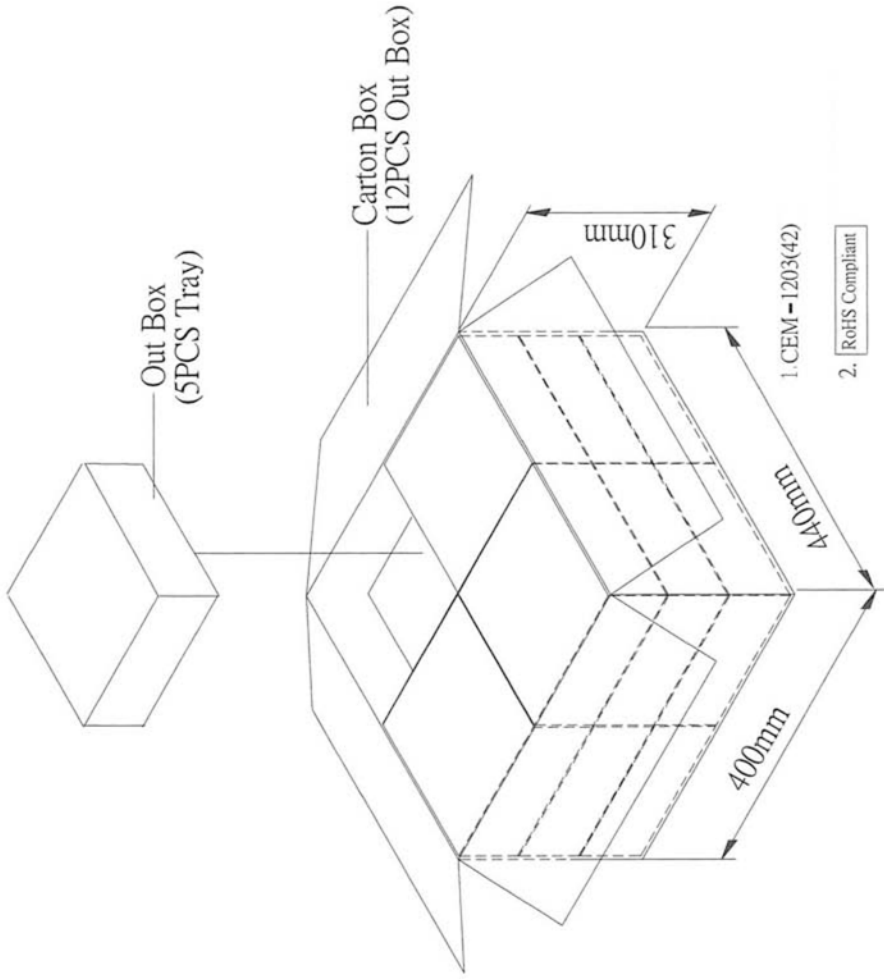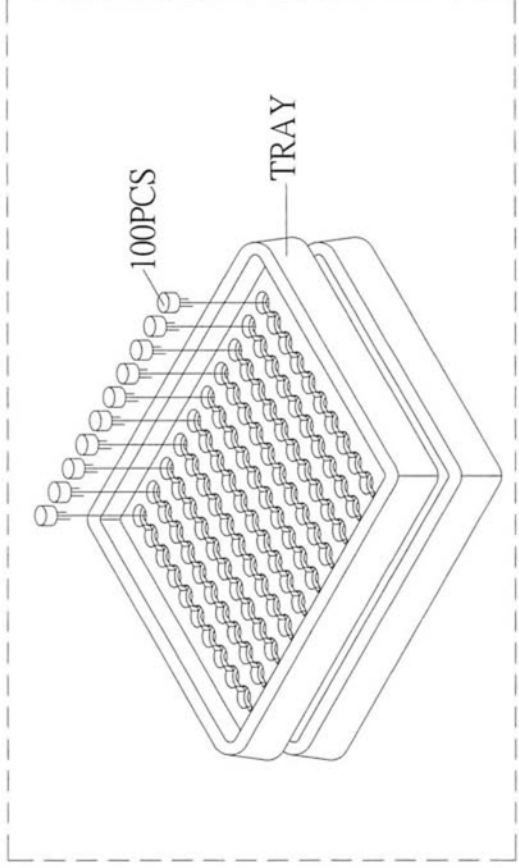
Tray	184mmx184mmx23mm	1x100PCS=100PCS
Out Box	200mmx190mmx100mm	5LAYERx100PCS=500PCS
Carton Box	440mmx400mmx310mm	500PCSx12=6000PCS

Appendix E-14: Mini Header Specifications

multicomp

PART NO.

2211S-08G

REVISIONS

ECN #	REV	DESCRIPTION	DRAWN	DATE	CHECKD	DATE	APPRVD	DATE
-	A	RELEASED	Veena	01/8/08	Suresh	01/8/08	G. C	15/8/08

Specifications:

Current rating	:	3A ac/dc.
Voltage rating	:	250V ac/dc.
Maximum contact resistance	:	20mΩ.
Minimum insulation resistance	:	1000MΩ.
Withstanding voltage	:	600V ac for one minute.
Operating temperature range	:	-40°C to +105°C.
Material and Finish:		
Insulator	:	High temperature plastic (PBT).
Insulator colour	:	Black.
Contact	:	Copper alloy.
Plated	:	Gold plated.
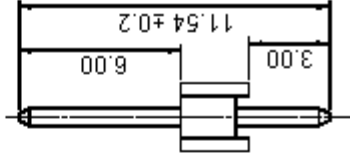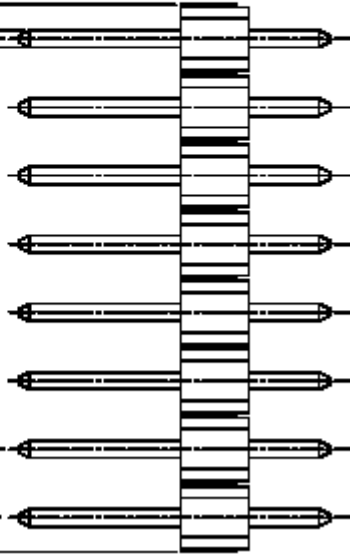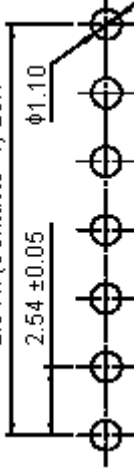
Dimensions : Millimetres

11.54 ±0.2

6.00

3.00

2.54 x Number of Positions ±0.35

0.64 Square

2.54 x (Number of Positions -1)

2.54 ±0.05

φ1.10

2.54 x (Contacts -1) ±0.1

2.54 ±0.05

PCB Layout

TOLERANCES:	DRAWN BY:	DATE:	DRAWING TITLE:		
UNLESS OTHERWISE SPECIFIED, DIMENSIONS ARE FOR REFERENCE PURPOSES ONLY.	Veena	01/08/08	Single Row Right Angle - 8 Way		
	CHECKED BY:	DATE:			
	Suresh	01/08/08	DWG NO. M10001419	SIZE A	DWG NO.
	APPROVED BY:	DATE:			ELECTRONIC FILE 2211S-XXG-F1_6_DWG
	G.Cook	15/08/08	SCALE: NTS	U.O.M.: mm	REV A

SHEET: 1 OF 2

REVISIONS

ECN #	REV	DESCRIPTION	DRAWN	DATE	CHECKD	DATE	APPRVD	DATE
-	A	RELEASED	Veena	01/8/08	Suresh	01/8/08	G. C	15/8/08

PART NO.

2211S-08G

Part Number Table

Description	Part Number
Header, 1 Row, Vertical, 8 Way	2211S-08G

Part Number Explanation:

2211   S   -   08   G

Right Angle   Number of Positions   Gold Plated
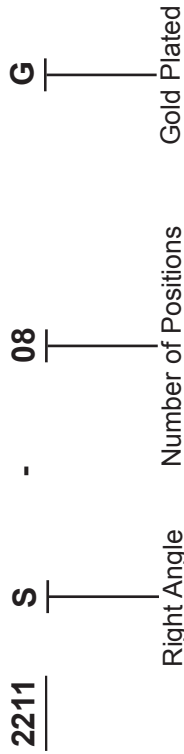
Number of Positions : 08 = 8 Positions.
Plated : G = Gold Plated.

DRAWING TITLE:

Single Row Right Angle - 8 Way

SIZE **A**	DWG NO. **M10001419**	
		ELECTRONIC FILE **2211S-XXG-F1_6_DWG**	REV **A**

SCALE: NTS   U.O.M.: mm   SHEET: 2   OF   2

TOLERANCES:	DRAWN BY:	DATE:
UNLESS OTHERWISE SPECIFIED, DIMENSIONS ARE FOR REFERENCE PURPOSES ONLY.	Veena	01/08/08
	CHECKED BY: Suresh	DATE: 01/08/08
	APPROVED BY: G.Cook	DATE: 15/08/08

Appendix E-15: Motor Driver Specifications

TOSHIBA

Toshiba Bi-CD Integrated Circuit   Silicon Monolithic

TB6612FNG

Driver IC for Dual DC motor

TB6612FNG is a driver IC for DC motor with output transistor in LD MOS structure with low ON-resistor. Two input signals, IN1 and IN2, can choose one of four modes such as CW, CCW, short brake, and stop mode.

Features

- Power supply voltage   VM=15V   Max.
- Output current   Iout=1.2A(ave) / 3.2A (peak)
- Output low ON resistor   0.5   (upper   lower Typ. @VM 5V)
- Standby (Power save) system
- CW/CCW/short brake/stop function modes
- Built-in thermal shutdown circuit and low voltage detecting circuit
- Small faced package SSOP24   0.65mm Lead pitch
- Response to Pb free packaging

SSOP24-P-300-0.65A

: 0.14 g (   )

---
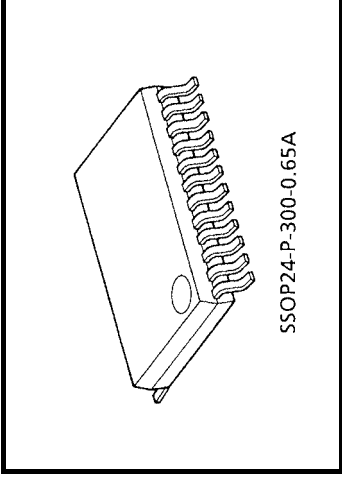
* This product has a MOS structure and is sensitive to electrostatic discharge. When handling this product, ensure that the environment is protected against electrostatic discharge by using an earth strap, a conductive mat and an ionizer. Ensure also that the ambient temperature and relative humidity are maintained at reasonable levels.

The TB6612FNG is a Pb-free product.
The following conditions apply to solderability:
*Solderability
1.   Use of Sn-37Pb solder bath
   *solder bath temperature = 230°C
   *dipping time = 5 seconds
   *number of times = once
   *use of R-type flux
2.   Use of Sn-3.0Ag-0.5Cu solder bath
   *solder bath temperature = 245°C
   *dipping time = 5 seconds

TOSHIBA

Block Diagram
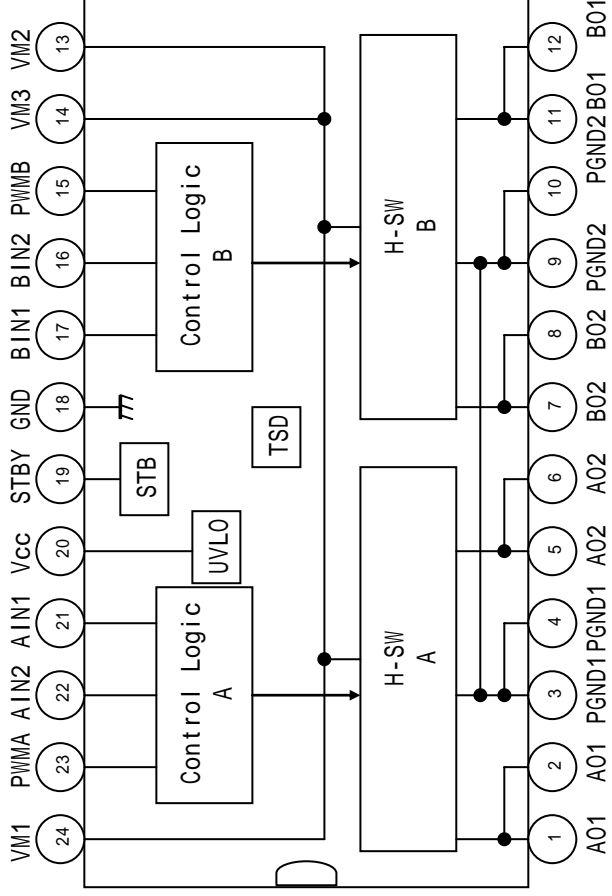
Pin Functions

Pin NO.	Symbol	I/O	Remarks
1	AO1	O	chA output1
2	AO1	O	chA output1
3	PGND1	—	Power GND 1
4	PGND1	—	Power GND 1
5	AO2	O	chA output2
6	AO2	O	chA output2
7	BO2	O	chB output2
8	BO2	O	chB output2
9	PGND2	—	Power GND 2
10	PGND2	—	Power GND 2
11	BO1	O	chB output1
12	BO1	O	chB output1
13	VM2	—	Motor supply 2.5V～13.5V
14	VM3	—	Motor supply 2.5V～13.5V
15	PWMB	I	chB PWM input / 200k pull-down at internal
16	BIN2	I	chB input2 / 200k pull-down at internal
17	BIN1	I	chB input1 / 200k pull-down at internal
18	GND	—	Small signal GND
19	STBY	I	"L"=standby / 200k pull-down at internal
20	Vcc	—	Small signal supply (2.7V～5.5V)
21	AIN1	I	chA input1 / 200k pull-down at internal
22	AIN2	I	chA input2 / 200k pull-down at internal
23	PWMA	I	chA PWM input / 200k pull-down at internal
24	VM1	—	Motor supply 2.5V～13.5V

TOSHIBA

Absolute Maximum Ratings (Ta = 25°C)

Characteristics	Symbol	Rating	Unit	Remarks
Supply voltage	VM	15	V	
Supply voltage	Vcc	6	V	
Input voltage	VIN	-0.2~6	V	IN1, IN2, STBY, PWM pins
Output voltage	Vout	15	V	O1, O2 pins
Output current	Iout	1.2	A	Per 1ch
Output current	Iout (peak)	2	A	tw≤20ms, Continuous pulse, Duty 20%
Output current	Iout (peak)	3.2	A	tw≤10ms, Single pulse
Power dissipation	PD	0.78	W	IC only
Power dissipation	PD	0.89	W	50×50 t=1.6(mm) Cu 40% in PCB mounting
Power dissipation	PD	1.36	W	76.2×114.3 t=1.6(mm) Cu 30% in PCB mounting
Operating temperature	Topr	-20~85		
Storage temperature	Tstg	-55~150		

Operating Range   Ta=-20~85

Characteristics	Symbol	Min	Typ.	Max	Unit	Remarks
Supply voltage	Vcc	2.7	3	5.5	V	
Supply voltage	VM	4.5	5	13.5	V	
Output current (1~5V)	Iout	---	---	1.0	A	VM≥5V
				0.4	A	5V>VM≥4.5V
Switching frequency	fPWM	---	---	100	kHz	

Input pin; IN1, IN2, PWM, STBY

Output pin; O1, O2
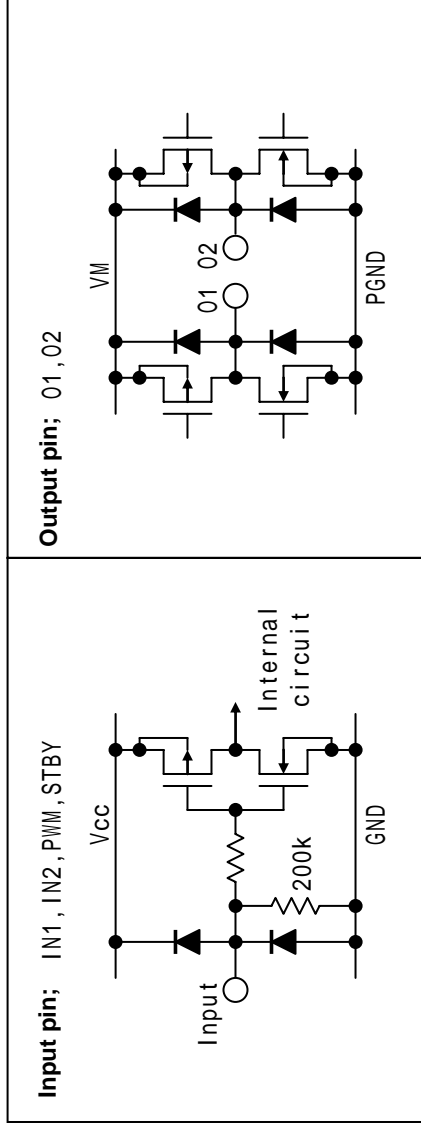
H-SW Control Function

Input				Output		
IN1	IN2	PWM	STBY	OUT1	OUT2	Mode
H	H	H/L	H	L	L	Short brake
L	H	H	H	L	H	CCW
L	H	L	H	L	L	Short brake
H	L	H	H	H	L	CW
H	L	L	H	L	L	Short brake
L	L	H	H	OFF (High impedance)		Stop
H/L	H/L	H/L	L	OFF (High impedance)		Standby

H-SW Operating Description

To prevent penetrating current, dead time t2 and t4 is provided in switching to each mode in the IC.

<ON>
t1

<OFF>
t2

<Short brake>
t3

<OFF>
t4

<ON>
t5

OUT1
Voltage wave

TOSHIBA
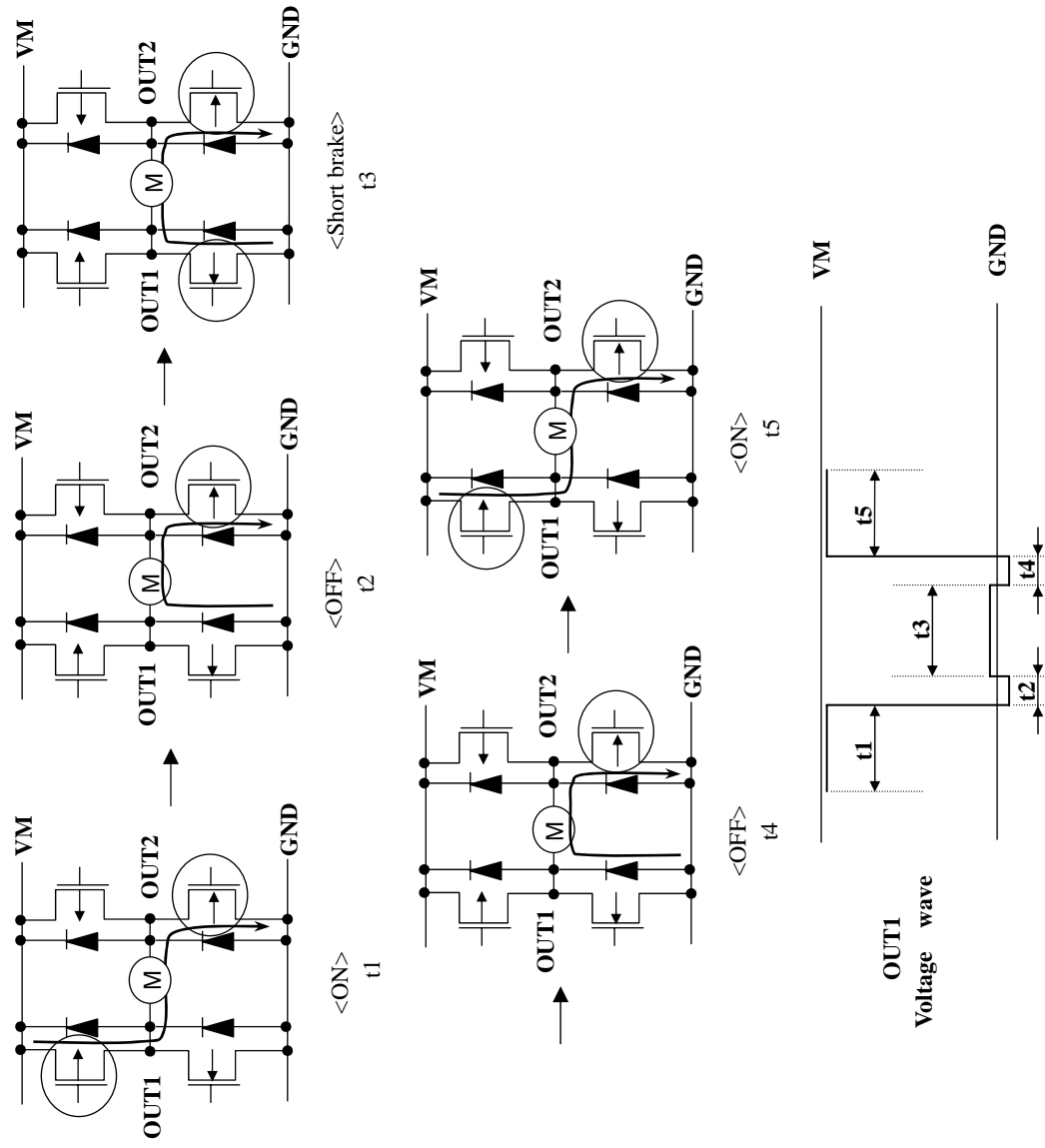
Electrical Characteristics (unless otherwise specified, Ta = 25°C, V$_{cc}$=3V, VM=5V)

Characteristics	Symbol		Test Condition	Min	Typ.	Max	Unit
Supply current	Icc(3V)		STBY=Vcc=3V, VM=5V	---	1.1	(1.8)	mA
	Icc(5.5V)		STBY=Vcc=5.5V, VM=5V	---	1.5	2.2	
	Icc(STB)		STBY=0V	---	---	1	μA
	IM(STB)		STBY=0V	---	---	1	
Control input voltage	VIH			Vcc×0.7	---	Vcc+0.2	V
	VIL			-0.2	---	Vcc×0.3	
Control input current	IIH		VIN=3V	5	15	25	μA
	IIL		VIN=0V	---	---	1	
Standby input voltage	VIH(STB)			Vcc×0.7	---	Vcc+0.2	V
	VIL(STB)			-0.2	---	Vcc×0.3	
Standby input current	IIH(STB)		VIN=3V	5	15	25	μA
	IIL(STB)		VIN=0V	---	---	1	
Output saturating voltage	Vsat(U+L)1		Io=1A,Vcc=VM=5V	---	0.5	(0.7)	V
	Vsat(U+L)2		Io=0.3A,Vcc=VM=5V		0.15	(0.21)	
Output leakage current	IL(U)		VM=Vout=15V	---	---	1	μA
	IL(L)		VM=15V,Vout=0V	-1	---	---	
Regenerative diode VF	VF(U)		IF=1A	---	1	1.1	V
	VF(L)			---	1	1.1	
Low voltage detecting voltage	UVLD		Designed value	---	1.9	---	V
Recovering voltage	UVLC		Designed value	---	2.2	---	
Response speed	tr		Designed value	---	24	---	ns
	tf			---	41	---	
	Dead time	H to L	Penetration protect time	---	(50)	---	
		L to H	Designed value	---	(230)	---	
Thermal shutdown circuit operating temperature	TSD		Designed value	---	175	---	
Thermal shutdown hysteresis	△TSD			---	20	---	

Target characteristics

PD - Ta

IC only Jθ a 16℃ /W

In boarding
PCB area 50× 50× 1.6mm
Cu area 40%

In boarding
PCB area 76.2× 114.3× 1.6mm
Cu area 30%

Iout - Duty

1ch driving

2ch driving

Ta=25 , IC only

TOSHIBA

TB6612FNG
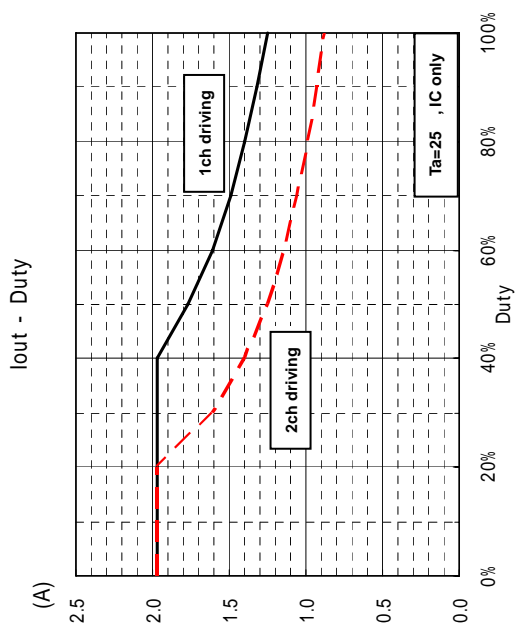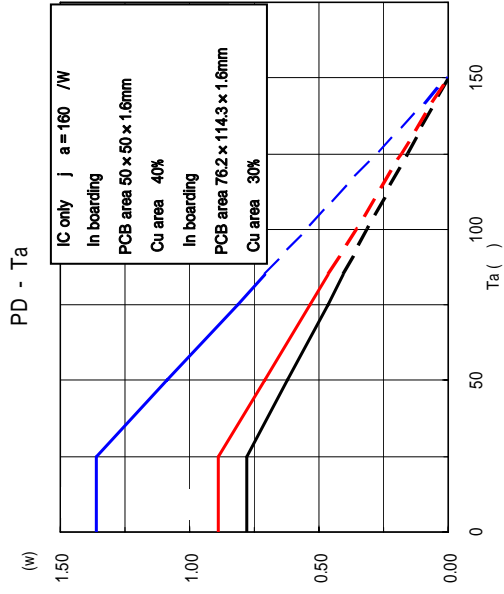
Typical Application Diagram
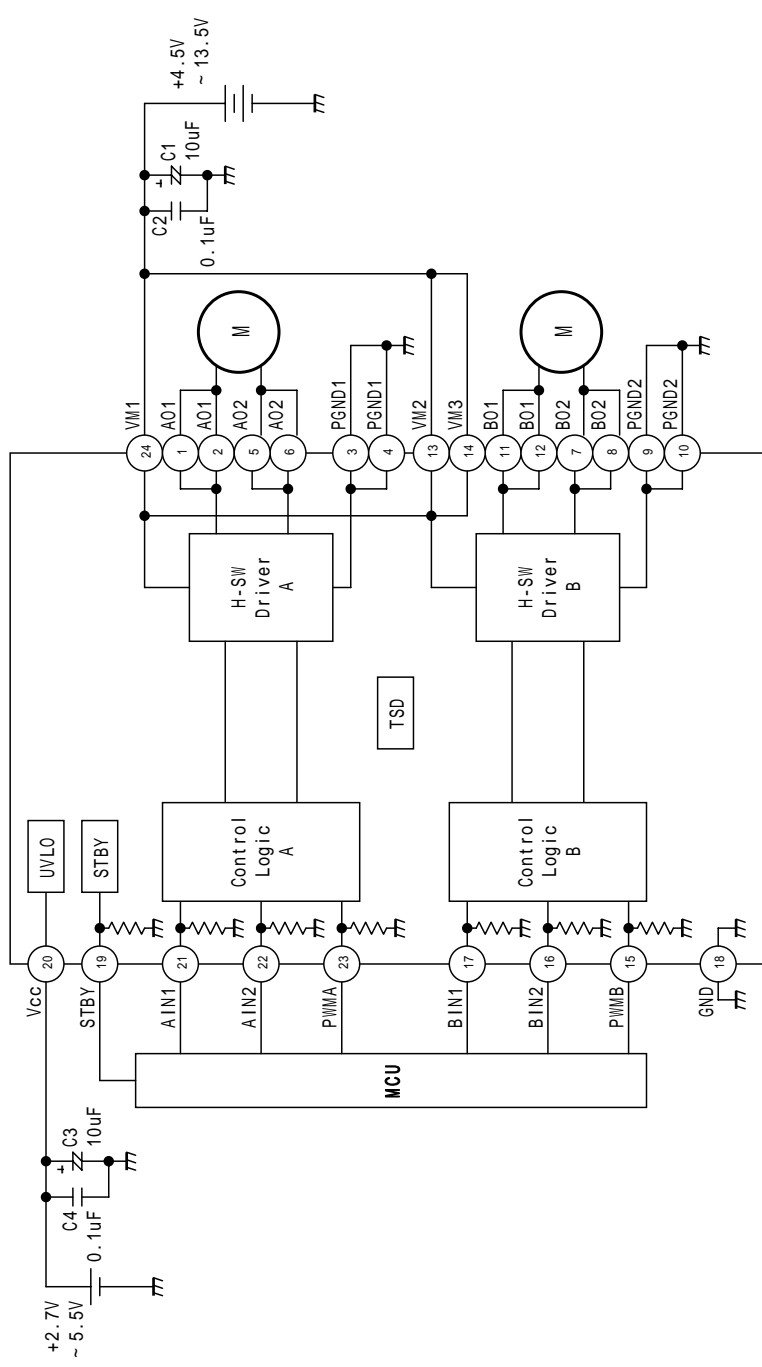
Note: Condensers for noise absorption (C1, C2, C3, and C4) should be connected as close as possible to the IC.

TOSHIBA

Package Dimennsions

SSOP24-P-300-0.65A

Unit : mm

7.6±0.3

5.6±0.2

13

12

24

1

0.22±0.1

0.65

0.325TYP

⊕ 0.13 Ⓜ

0.45±0.2

0.15 +0.1 −0.05

8.3MAX

7.8±0.2

1.6MAX

1.2±0.2

0.1 +0.1 −0.05
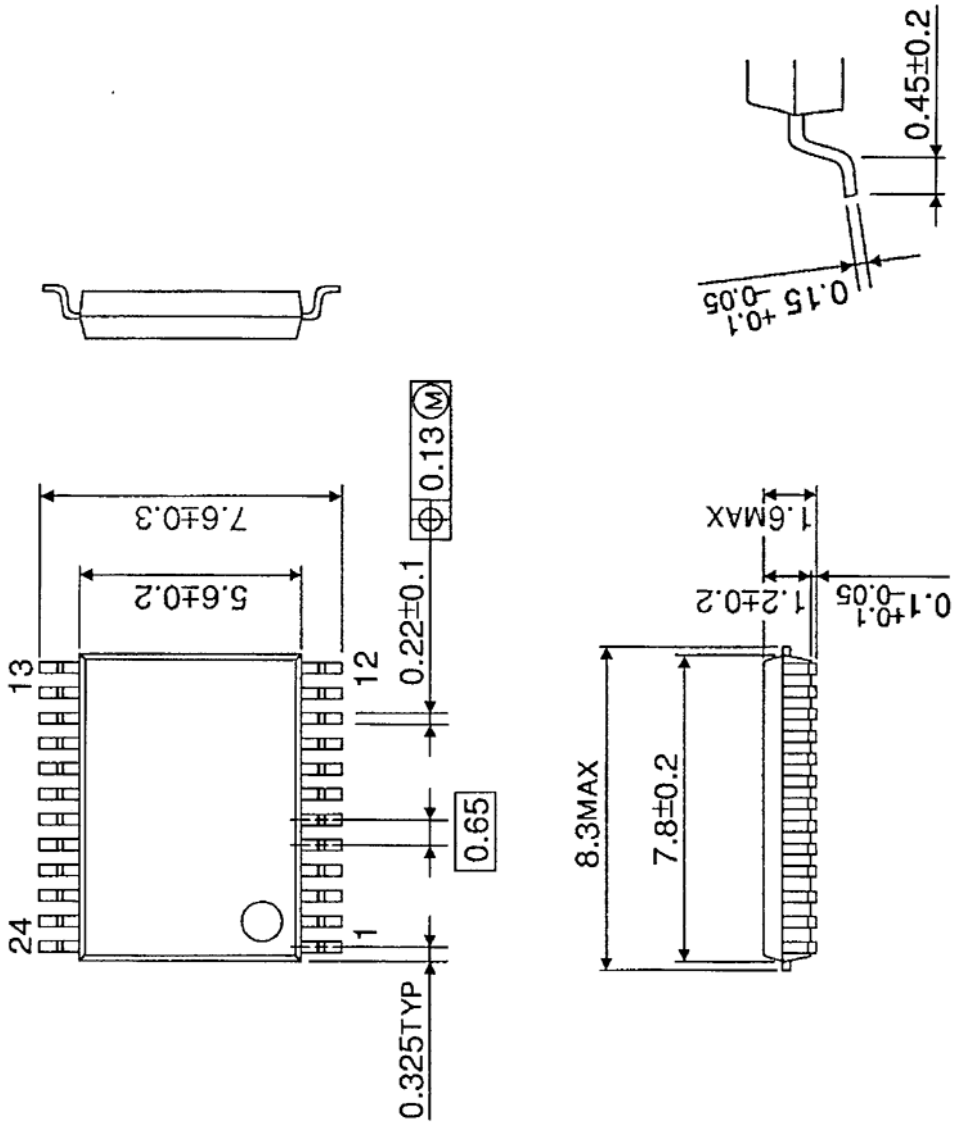
Weght: 0.14 g (typ)

Notes on Contents

1. Block Diagrams
Some of the functional blocks, circuits, or constants in the block diagram may be omitted or simplified for explanatory purposes.

2. Equivalent Circuits
The equivalent circuit diagrams may be simplified or some parts of them may be omitted for explanatory purposes.

3. Timing Charts
Timing charts may be simplified for explanatory purposes.

4. Application Circuits
The application circuits shown in this document are provided for reference purposes only. Thorough evaluation is required, especially at the mass production design stage.
Toshiba does not grant any license to any industrial property rights by providing these examples of application circuits.

5. Test Circuits
Components in the test circuits are used only to obtain and confirm the device characteristics. These components and circuits are not guaranteed to prevent malfunction or failure from occurring in the application equipment.

IC Usage Considerations
Notes on handling of ICs

[1] The absolute maximum ratings of a semiconductor device are a set of ratings that must not be exceeded, even for a moment. Do not exceed any of these ratings.
Exceeding the rating(s) may cause the device breakdown, damage or deterioration, and may result injury by explosion or combustion.

[2] Use an appropriate power supply fuse to ensure that a large current does not continuously flow in case of over current and/or IC failure. The IC will fully break down when used under conditions that exceed its absolute maximum ratings, when the wiring is routed improperly or when an abnormal pulse noise occurs from the wiring or load, causing a large current to continuously flow and the breakdown can lead smoke or ignition. To minimize the effects of the flow of a large current in case of breakdown, appropriate settings, such as fuse capacity, fusing time and insertion circuit location, are required.

[3] If your design includes an inductive load such as a motor coil, incorporate a protection circuit into the design to prevent device malfunction or breakdown caused by the current resulting from the inrush current at power ON or the negative current resulting from the back electromotive force at power OFF. IC breakdown may cause injury, smoke or ignition.
Use a stable power supply with ICs with built-in protection functions. If the power supply is unstable, the protection function may not operate, causing IC breakdown. IC breakdown may cause injury, smoke or ignition.

[4] Do not insert devices in the wrong orientation or incorrectly.
Make sure that the positive and negative terminals of power supplies are connected properly.
Otherwise, the current or power consumption may exceed the absolute maximum rating, and exceeding the rating(s) may cause the device breakdown, damage or deterioration, and may result injury by explosion or combustion.
In addition, do not use any device that is applied the current with inserting in the wrong orientation or incorrectly even just one time.

Points to remember on handling of ICs

(1) Thermal Shutdown Circuit

Thermal shutdown circuits do not necessarily protect ICs under all circumstances. If the thermal shutdown circuits operate against the over temperature, clear the heat generation status immediately.

Depending on the method of use and usage conditions, such as exceeding absolute maximum ratings can cause the thermal shutdown circuit to not operate properly or IC breakdown before operation.

(2) Heat Radiation Design

In using an IC with large current flow such as power amp, regulator or driver, please design the device so that heat is appropriately radiated, not to exceed the specified junction temperature (T_J) at any time and condition. These ICs generate heat even during normal use. An inadequate IC heat radiation design can lead to decrease in IC life, deterioration of IC characteristics or IC breakdown. In addition, please design the device taking into considerate the effect of IC heat radiation with peripheral components.

(3) Back-EMF

When a motor rotates in the reverse direction, stops or slows down abruptly, a current flow back to the motor's power supply due to the effect of back-EMF. If the current sink capability of the power supply is small, the device's motor power supply and output pins might be exposed to conditions beyond maximum ratings. To avoid this problem, take the effect of back-EMF into consideration in system design.

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070122EBA_R6

- The information contained herein is subject to change without notice. 021023_D

- TOSHIBA is continually working to improve the quality and reliability of its products. Nevertheless, semiconductor devices in general can malfunction or fail due to their inherent electrical sensitivity and vulnerability to physical stress. It is the responsibility of the buyer, when utilizing TOSHIBA products, to comply with the standards of safety in making a safe design for the entire system, and to avoid situations in which a malfunction or failure of such TOSHIBA products could cause loss of human life, bodily injury or damage to property.
  In developing your designs, please ensure that TOSHIBA products are used within specified operating ranges as set forth in the most recent TOSHIBA products specifications. Also, please keep in mind the precautions and conditions set forth in the "Handling Guide for Semiconductor Devices," or "TOSHIBA Semiconductor Reliability Handbook" etc. 021023_A

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- The products described in this document are subject to foreign exchange and foreign trade control laws. 060925_E

Appendix E-16: Card Connector MicroSD Specifications

3M™ Card Connector microSD™

Normal Polarization, Push-Push, Surfacemount

2900 Series
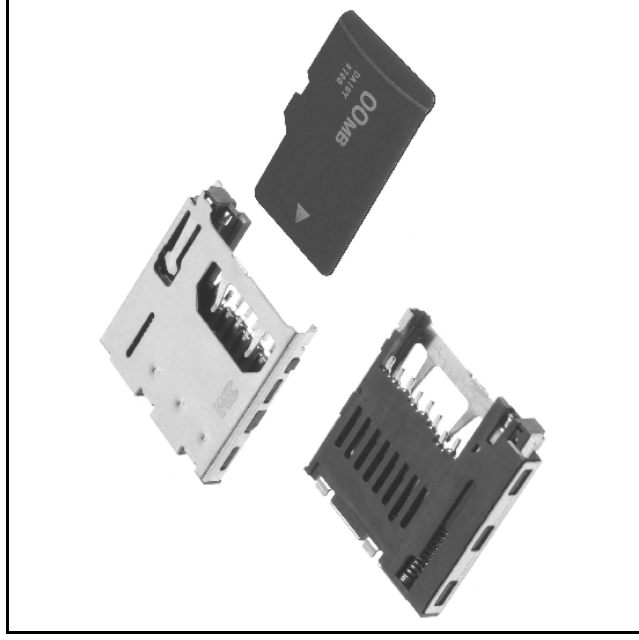
- 1.85 mm low profile height
- Small size of 14.0 mm × 14.2 mm footprint
- Smooth push–push eject mechanism
- Card is retained when mated
- Card polarization
- Will accept a card thickness of 0.8 mm
- Mechanism prevents card from flying out upon ejection
- Card detection indicator
- Metal–shielded cover
- microSD™ specification compliant
- RoHS* compliant

Physical

Insulation Material:	High Temperature Thermoplastic
Flammability:	UL 94V-0
Color:	Black
Contact Material:	Copper Alloy
Plating	
Underplating:	50 µ″ [1.27 µm] min. Nickel
Wiping Area:	15 µ″ [0.38 µm] min. Gold
Termination Area:	3 µ″ [0.076 µm] max. Gold flash
Cover Material:	Stainless Steel
Underplating:	40 µ″ [1.02 µm] min. Nickel
Solder Area:	Gold flash
Lock Pin and Link Pin Material:	Stainless Steel
Spring Material:	Copper Alloy
Plating:	30 µ″ [0.762 µm] min. Nickel
Marking:	3M Logo

Electrical

Current Rating:	0.5 A
Insulation Resistance:	100 MΩ min.
Withstanding Voltage:	500 VAC for one minute

Environmental

Operating Temperature:	-25°C to +85°C
Storage Temperature:	-40°C to +85°C
Process Rating:	Maximum 250°C, with 40 seconds over 230°C

For technical, sales or ordering information call
800-225-5373

3M

Interconnect Solutions
http://www.3M.com/interconnects/

3M™ Card Connector microSD™

Normal Polarization, Push-Push, Surfacemount

2900 Series

CIRCUIT SCHEMATIC
CARD DETECT INDICATION

(PIN NO. 2)

(PIN NO. 2)

PAD NO. 2 (CARD)

PAD NO. 2 (CARD)

PAD NO. 2 (CARD)

WITHOUT CARD

CARD INSERTED

0.60 [0.024] FLAT AREA

1.30 [0.051]

8X 0.80 [0.03]

REF 13.90 [0.547]

8X 0.35 [0.014]

0.60 [0.024] FLAT AREA

2X 1.20 [0.047]

7.15 [0.282]

0.85 [0.033]

1.30 [0.051]

REF 12.90 [0.508]

2X 0.60 [0.024] FLAT AREA

1.45 [0.057]

BOTTOM SIDE VIEW

7X PITCH 1.10 [0.043]

1.24 [0.049]

7.70 [0.303]

1.85 [0.073]

5.74 [0.226]

14.20 [0.559]

14.00 [0.551]

REF 5.40 [0.213] *

REF 1.60 [0.063] **

0.70 [0.028] ***

15.20 [0.598]

TOP SIDE VIEW

3M

Notes:
*   Card fully ejected position
**   Card in locked position
***   Card fully pushed in

Tolerance Unless Noted

	.0	.00	.000
mm	±.25	±.15	±.05

mm
[Inch]

[ ] Dimensions for Reference Only

3M is a trademark of 3M Company.

For technical, sales or ordering information call
800-225-5373

TS-2197-02
Sheet 2 of 3

Interconnect Solutions
http://www.3M.com/interconnects/

3M™ Card Connector microSD™

Normal Polarization, Push-Push, Surfacemount
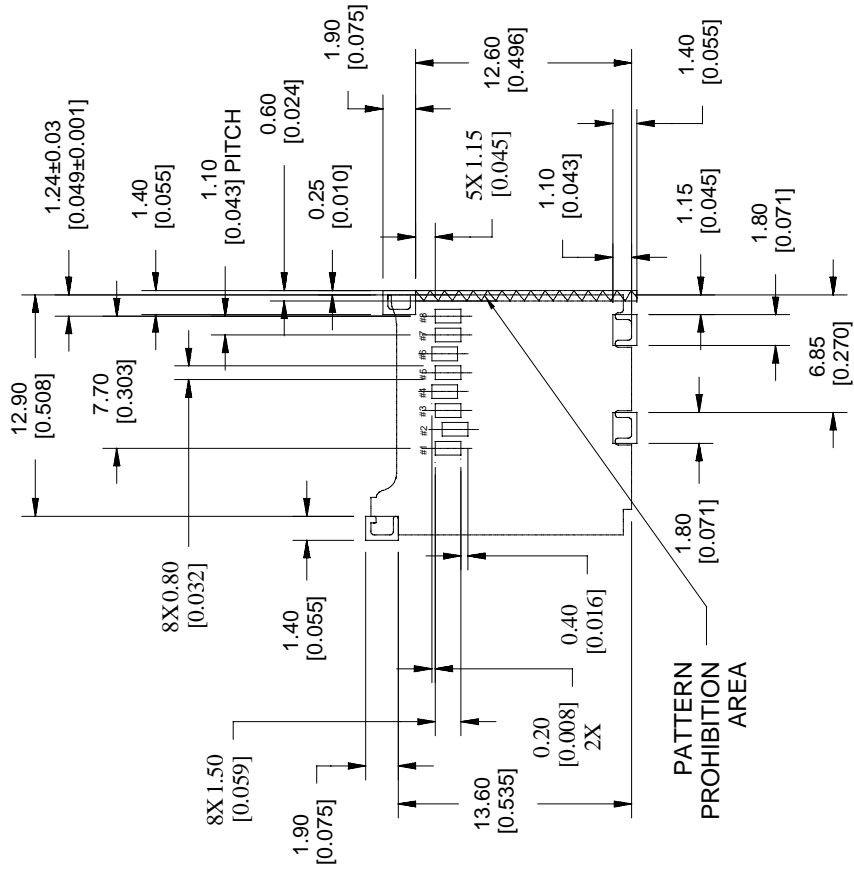
2900 Series

RECOMMENDED
PCB PATTERN LAYOUT ± 0.05 [0.002]

12.90 [0.508]
7.70 [0.303]
8X0.80 [0.032]
1.40 [0.055]
8X1.50 [0.059]
1.90 [0.075]
13.60 [0.535]
0.20 [0.008] 2X
0.40 [0.016]
PATTERN PROHIBITION AREA
1.80 [0.071]

1.24±0.03 [0.049±0.001]
1.40 [0.055]
1.10 [0.043] PITCH
0.60 [0.024]
0.25 [0.010]
1.90 [0.075]
5X 1.15 [0.045]
12.60 [0.496]
1.40 [0.055]
1.10 [0.043]
1.15 [0.045]
1.80 [0.071]
6.85 [0.270]

Ordering Information

29 - XX - X X XX - XX

2900 Series:

Contact Count = 08

Ejection Type:
0 = Push-Push

Card Polarization, Ejection Type, Overall Heght
5 = Normal, Spring-loaded Push Eject, 1.85 mm Height

Packing Options:
WB = 24 mm Tape and Reel

Plating Suffix:
MG = 15 µ" min. Au on contact
Au Flash on Tail (50 µ" Ni min. underplating) (RoHS* compliant)

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Post-Consumer Fiber

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Minimum 10%
Post-Consumer Fiber

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Appendix E-17: Rotary Encoder Specifications

Encoders

ALPS 8-DIRECTIONAL SWITCH AND ENCODER WITH CENTER PUSH

This product is RoHS compliant.

- Single shaft has 8 directional switch, push center, and encoder operations (15 pulses, 30 clicks)
- Extremely low profile
- User-friendly 8-directional switch
- Rating: 10mA @ 5VDC
- Operating life (Switch/Encoder): 50,000 cycles/15,000 cycles
- Switch operating angle: 12° max.
- Control sound, cursors, and switching modes in car audio and navigation systems
- Input controls for home electronics

RoHS Compliant

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alps Part No.	Fig.	Switch OF	Detent Torque Encoder	Encoder No. of			Price Each			
					Detents	Pulses	1	25	50	100	
688-RKJXT1E12001	RKJXT1E12001	A	40±25 mN·m	15±8 mN·m	30	15					

ALPS RK097 SERIES: 9MM METAL SHAFT ENCODERS

- Rating: 10mA @ 12 VDC
- Rotational life: 30,000 cycles
- Incremental type
- Detent torque: 10+/-8mN·m
- Applications include controls for AV devices, level controls for communication devices

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alps Part No.	Fig.	Push-On Switch	Travel of Push-On Switch	Res.	No. of Detents	Price Each			
							1	25	50	100
688-RK09710ELC09	RK09710ELC09	B	With	0.5mm	9	18				
688-RK09710ELC07	RK09710ELC07	B	With	0.5mm	15	30				
688-RK09710ELC0B	RK09710ELC0B	C	With	1.5mm	15	30				

ⓐ ROTARY ENCODERS

ALPHA

RE701 Series: 7mm Metal Shaft Rotary Encoders ■■■ NEW FROM SUPPLIER
- Contact Resistance: 100Ω Max
- Rotational Life: 50,000 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rating	No. of Pulses	No. of Detents	Push-on Switch	Price Each			
							1	25	50	100
318-ENC70121F-10PS	RE701F-21B1-15F-10P-601	A	1mA/5VDC	10	20	Yes				
318-ENC70120F-10P	RE701F-20B1-15F-10P-601	B	1mA/5VDC	10	20	No				

RE111 Series: 11mm Metal Shaft Rotary Encoders
- Rating: 10mA /5VDC (1mA Min.)
- Insulation Resistance: 300VDC 100MΩ min.

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rotational Torque	No. of Pulses	No. of Detents	Push-on Switch	Price Each			
							1	25	50	100
318-ENC111F-20P	RE111F-20B3-20F-20P	C	30~200gf.cm	20	20	No				
318-ENC111F-20PS	RE111F-21B3-20F-20P	C	30~200gf.cm	20	20	Yes				

RE11 Series: 11mm Metal Shaft Rotary Encoders ■■■ NEW FROM SUPPLIER
- Rating: 10mA /5VDC (1mA Min.)
- Contact Resistance: 100Ω Max.
- ✱ Self-Return Type
- Shaft Push-Pull Strength: 10kgf
- Rotational Life: 15,000 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Dim. (mm) L	F	No. of Pulses	No. of Detents	Push-on Switch	Price Each			
								1	25	50	100
318-ENC1110F-15P	RE11LF-40B1N-10F-15P	D	10	3.7	15	30	No				
318-ENC1112F-15PS	RE11LF-41B1N-12F-15P	D	12	5.0	15	30	Yes				
✱318-ENC1120F-15PS	RE11RF-41B3N-20F-0P	E	-	-	-	-	Yes				

RE130 Series: 12mm Insulated Shaft Rotary Encoders
- Insulation Resistance: 50VDC 10MΩ min
- Rotational Life: 30,000 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rating	No. of Pulses	No. of Detents	Push-on Switch	Price Each			
							1	25	50	100
318-ENC13020F-12P	RE130F-40-20F-12P	F	1mA/10VDC	12	12	No				
318-ENC13015F-12P	RE130F-40E3-15F-12P-26	G	1mA/10VDC	12	12	No				
318-ENC13017SF-12PS	RE130F-41-175F-12P	H	1mA/10VDC	12	12	Yes				

RE160 SERIES: 16MM INSULATED SHAFT ROTARY ENCODER
- Insulation Resistance: 50VDC 10MΩ min
- Rotational Life: 50,000 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rating	No. of Pulses	No. of Detents	Push-on Switch	Price Each			
							1	25	50	100
318-ENC160F-24P	RE160F-40E3-20A-24P	I	0.5mA/5VDC	24	24	No				

RE170 SERIES: 17MM GRAY CODED ROTARY ENCODER ■■■ NEW FROM SUPPLIER
- Insulation Resistance: 100VDC 100MΩ min
- Rotational Torque: 200~500 gf.cm
- Rotational Life: 10,000 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rating	No. of Detents	Push-on Switch	Price Each			
						1	25	50	100
318-ENC170248F-15	RE170F-40E3-248F-4G-15	J	0.1A/12VDC	15	No				

RE200 SERIES: 20MM GRAY & BINARY CODED ROTARY ENCODERS
- Insulation Resistance: 100±10VDC 10MΩ min
- Rotational Torque: 50~350 gf.cm
- Rotational Life: 50,000±2500 cycles

For quantities of 200 and up, call for quote.

MOUSER STOCK NO.	Alpha Part No.	Fig.	Rating	No. of Detents	Output Code	Price Each			
						1	25	50	100
318-ENC200120F-4B	RE2001F-40E2-20F-4B	K	10mA/10VDC	16	Binary				
318-ENC200155F-4G	RE200F-40E2-155F-4G	L	10mA/10VDC	16	Gray				
318-ENC20020F-4B	RE200F-20E2-20F-4B	M	10mA/10VDC	16	Binary				

Ⓜ Mountain Switch ROTARY ENCODERS WITH CENTER PUSH SWITCH ■■■ NEW FROM SUPPLIER

RoHS Compliant

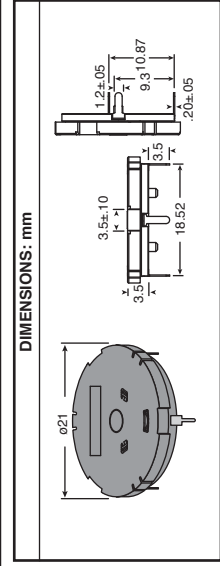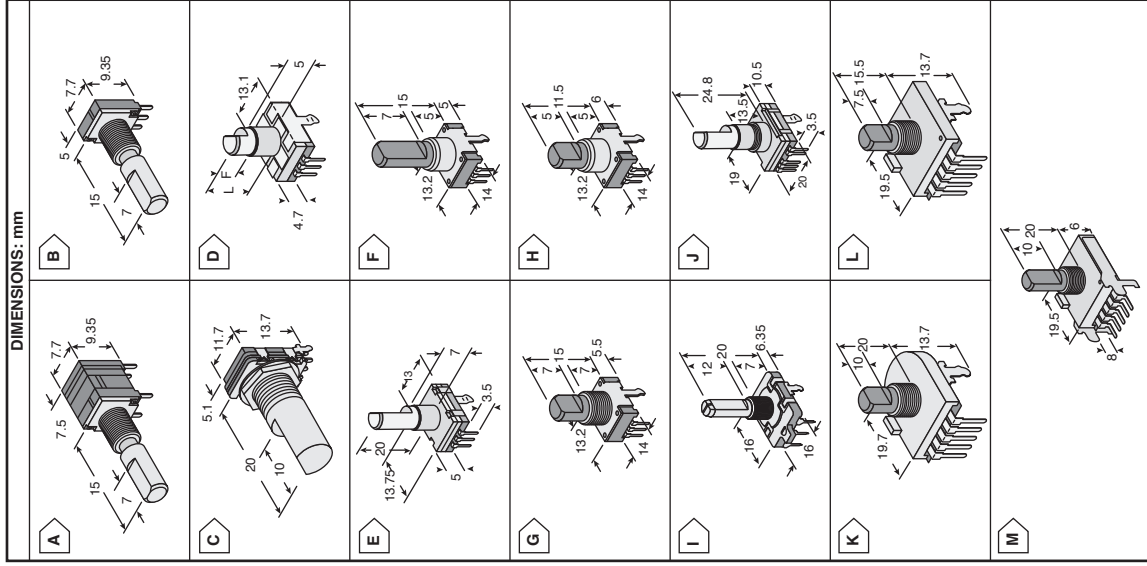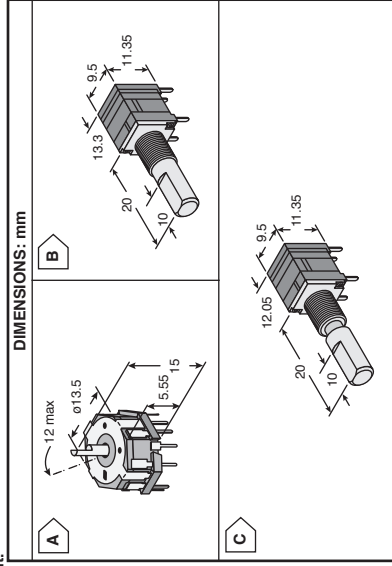**Electrical Specifications:**
- Contact Rating: 1mA/10VDC
- Contact Resistance: Max. 300Ω (Initial)
- Insulation Resistance: Min. 100MΩ 100VDC

For quantities of 500 and up, call for quote.

MOUSER STOCK NO.	No. of Pulses	No. of Detents	Operating Force Rotation	Push	Holding Hooks	Price Each			
						1	25	50	100
101-115-6191-EV	12	24	50gf-cm (Max)	220±40 gf	3				
101-115-6110-EV	12	24	50gf-cm (Max)	220±40 gf	6				

DIMENSIONS: mm

DIMENSIONS: mm

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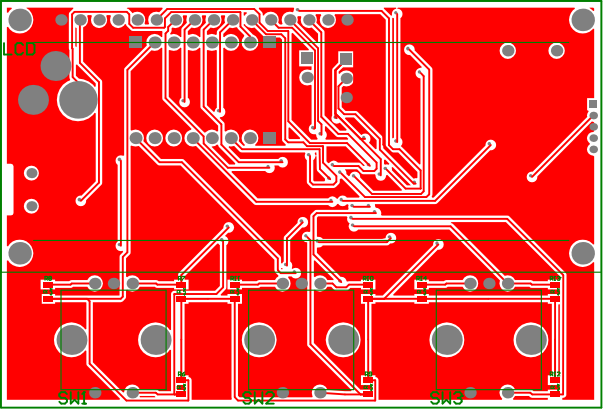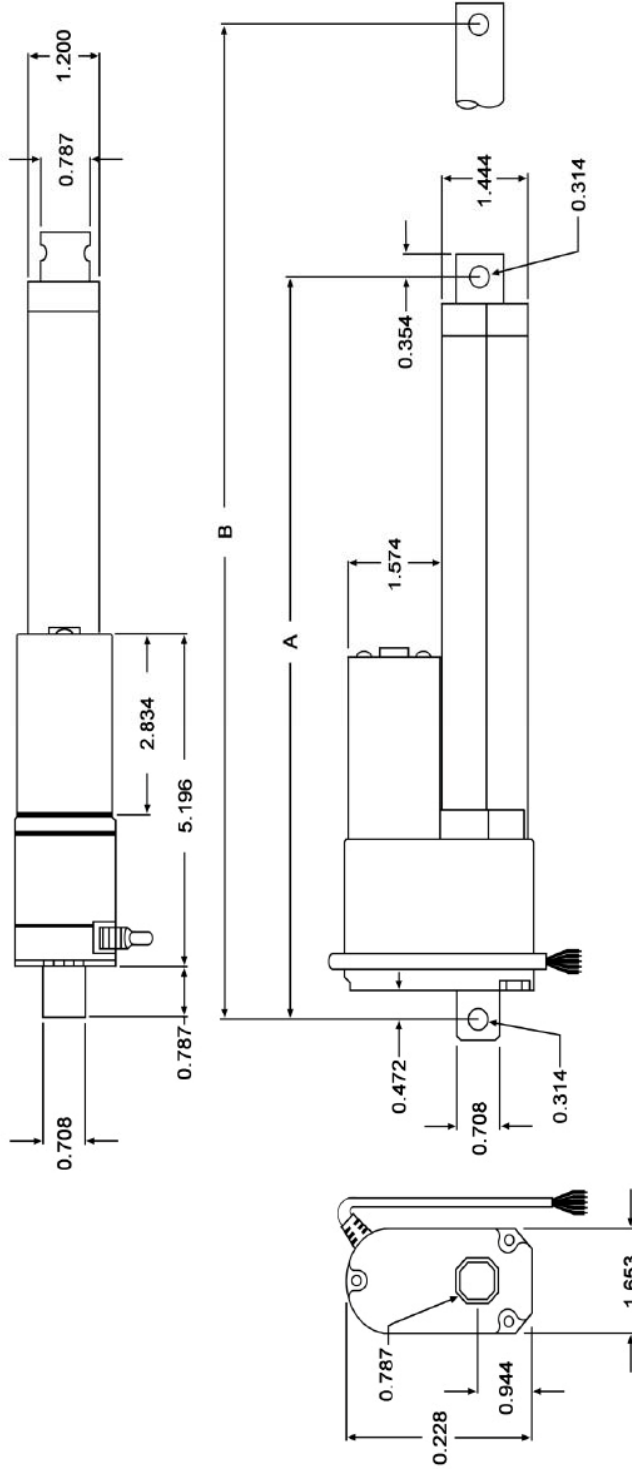**Appendix E-18: Circuit Board Layout**

LCD

SW1   SW2   SW3

Appendix F: Engineering Drawings

APPENDIX F: DRAWINGS

Linear Actuator
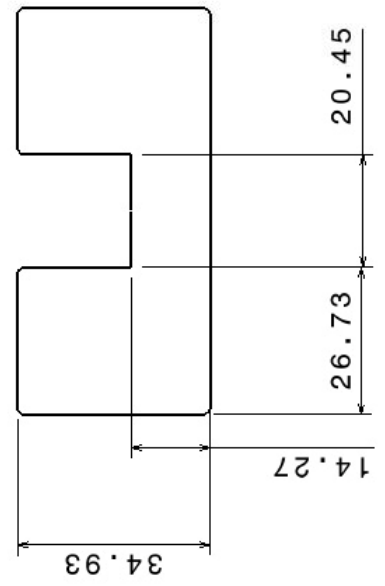
12 VDC Linear Actuator 107 lbs.

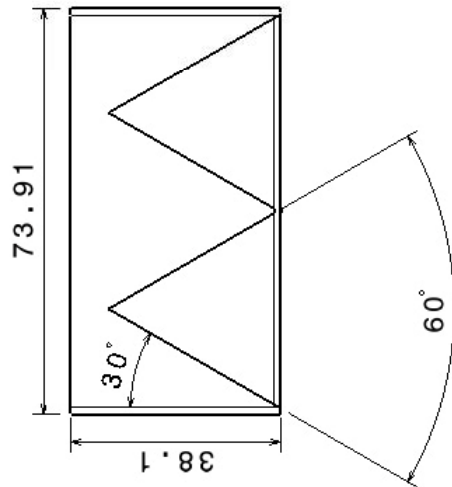| Item Number	5-1577-2	5-1577-4	5-1577-6	5-1577-8	5-1577-10	5-1577-12
Stroke	1.96	3.93	5.90	7.87	9.84	11.81
A	7.68	9.69	11.69	13.70	15.71	17.72
B	9.65	13.62	17.60	21.57	25.55	29.53

SURPLUS CENTER
800-488-3407
www.surpluscenter.com
1015 West "O" St.
LINCOLN, NE 68528

All dimensions are in inches

Slider
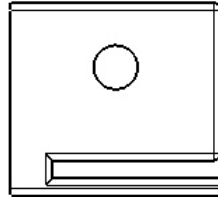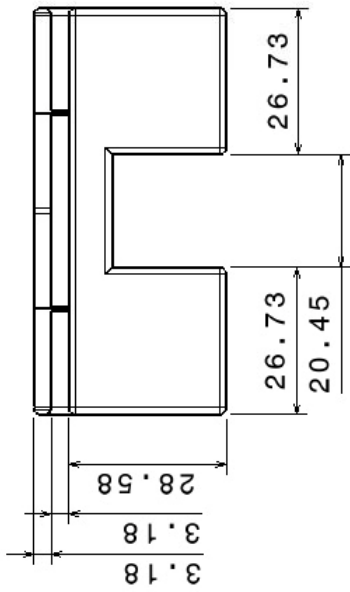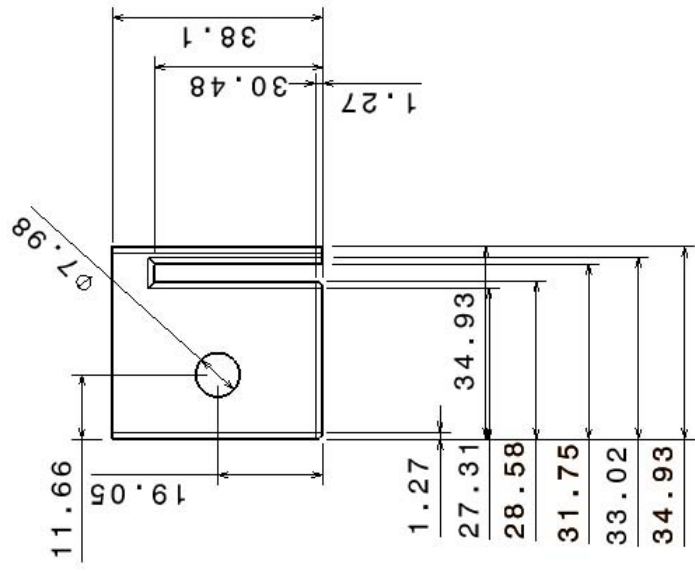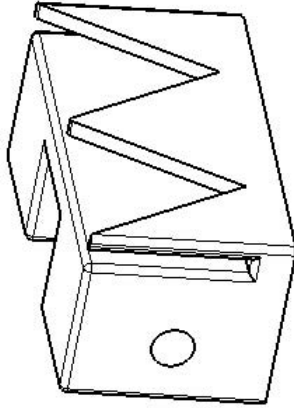
Holding Block

ALL DIMENSIONS IN MM

38.1

31.12

6.35

127

88.9

2.54

90°

31.12

127

71.86

45.72

14.61

62.88

6.43

Brace

R3.18

R3.18

15.88

9.53

88.9

19.05

R2.54

17.78

11.43

6.35

25.4

76.2

R6.35

25.4

31.75

127
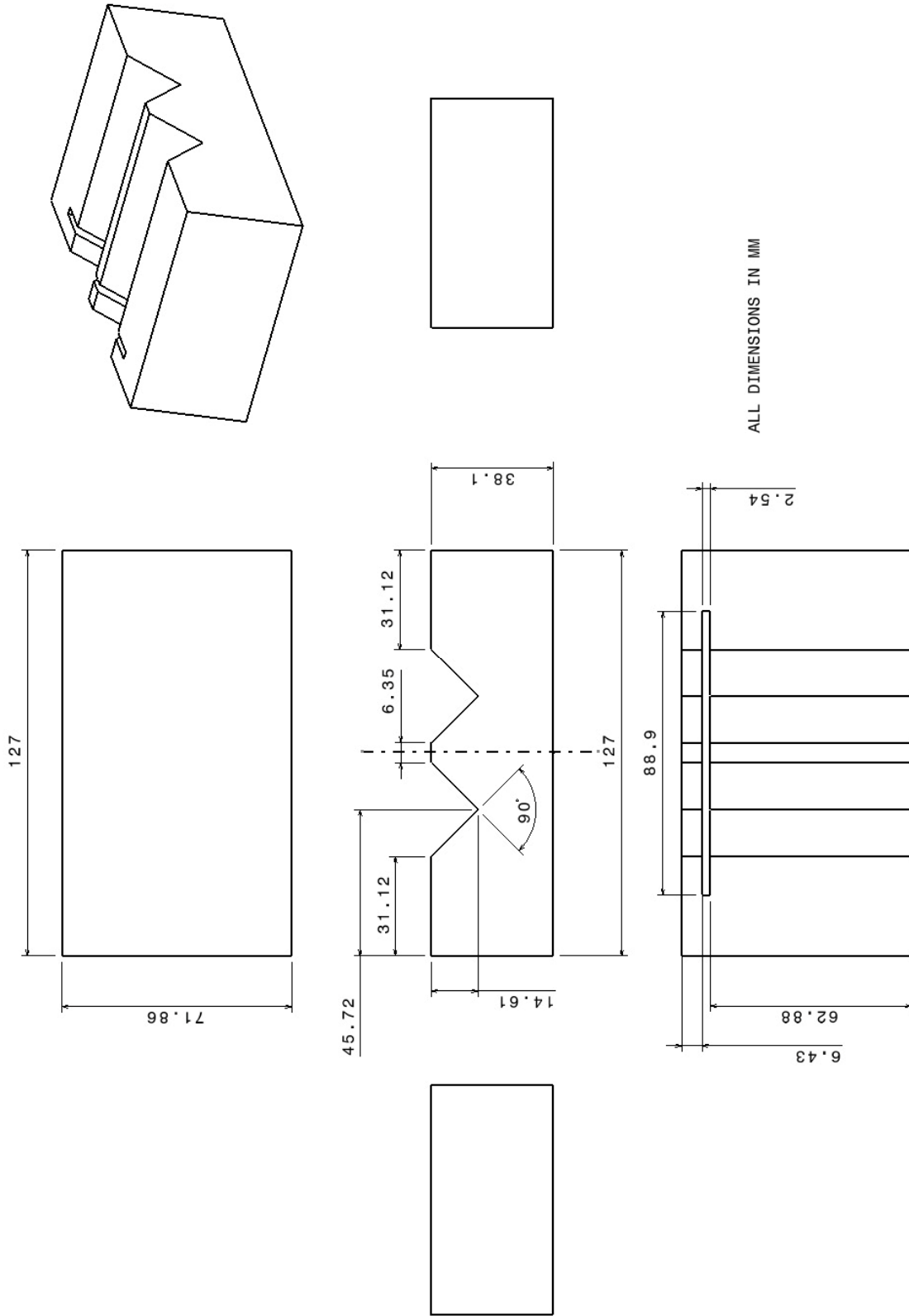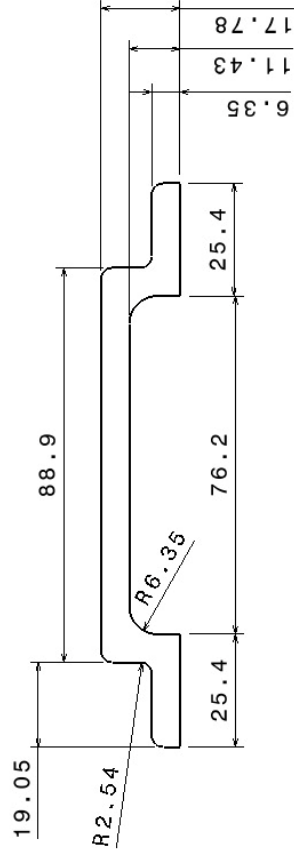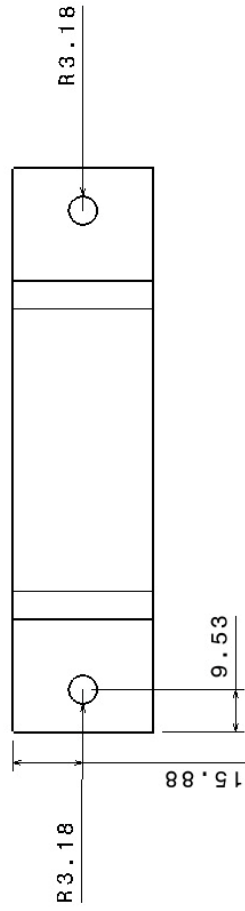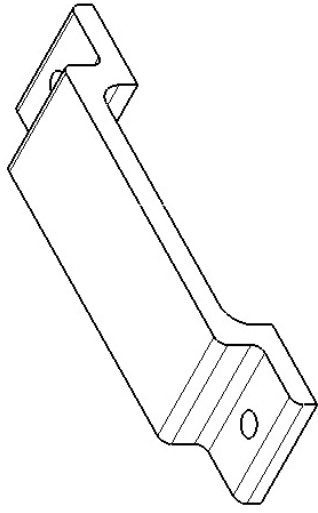
ALL DIMENSIONS IN MM
ALL UN-DIMENSIONED FILLETS ARE R2.54
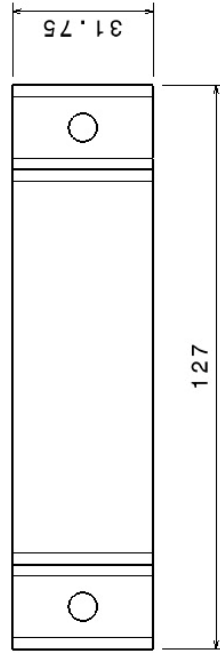
Appendix G: Instruction Manual

Figure 1.

A→ B C
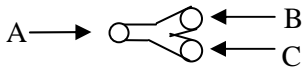
Figure 2. One-way valve orientation.

Front Side   Back Side

Figure 3.

A→ D E

Figure 4.

A→ F G

Figure 5.

A→ H I

Figure 6.

A→ J K

Figure 7.

A→ L M

Figure 8.

A→ N O P Q

Figure 9.

Figure 10.

Figure 11.

Figure 12.

Figure 13.

14. Connect port V (refer to Figure 12 for the location of port V) to a fresh blood bag of the required blood type. (Refer to Figure 13.)
15. Connect port W (refer to Figure 13 for the location of port W) to a empty IV bag. (Refer to Figure 14.)
16. Connect the fresh blood bag onto an IV stand so that the fresh blood bag is at a lower level than the neonate.
17. Connect the empty IV bag onto an IV stand so that the empty IV bag is at a higher level than the neonate.
18. Obtain two syringes of the same volume and brand (between 20 and 25 mL).
19. Place the syringes in the syringe holder with the lip of the syringe hitting the side of the syringe holder. (Refer to Figure 15.)
20. Plug the power cord into the DC power jack. (Refer to Figure 16.)
21. Plug the power cord into a wall outlet.
22. Rotate encoder 1 (refer to Figure 17), located on the front side of the device, so that the slider (refer to Figure 18) is translated to the position just before it hits the syringe holder.
    a. Rotate the encoder clockwise to translate the slider to the left.
    b. Rotate the encoder counterclockwise to translate the slider to the right.
23. Place the syringe plunger snuggly into the slider. (Refer to Figure 19.)
24. With the syringe plunger still snuggly fit into the slider, extend plunger so that the lip of the plunger is flush with the side of the syringe holder.
25. Create a snug fit between the plate on the side of the syringe holder and the syringe lip by tightening the wing nuts. (Refer Figure 20.)
26. Place the brace of the syringe holder of the syringes.
27. Lock the syringes in place by tightening the nut above the brace. (Create a snug fit between the brace and the syringe holder.) (Refer to Figure 21.)
28. Rotate encoder 1(refer to Figure 17) to the right until the plunger of the syringe is fully extended. (Do not extend the plunger further than the syringe can handle.)
29. Press encoder 1 (refer to Figure 17) to set the position of the syringe when its plunger is fully extended.

Figure 14.

Figure 15.

Figure 16.

Power Adapter   DC Power Jack

Figure 17.
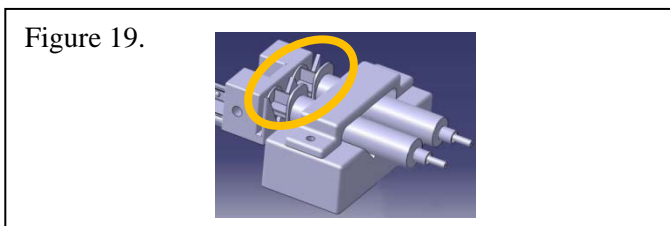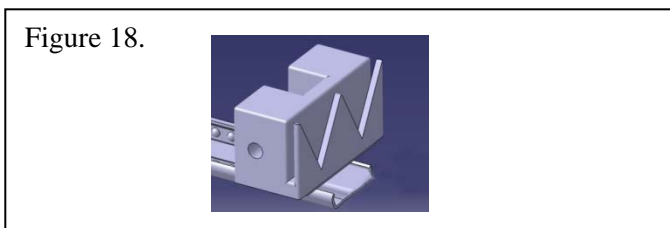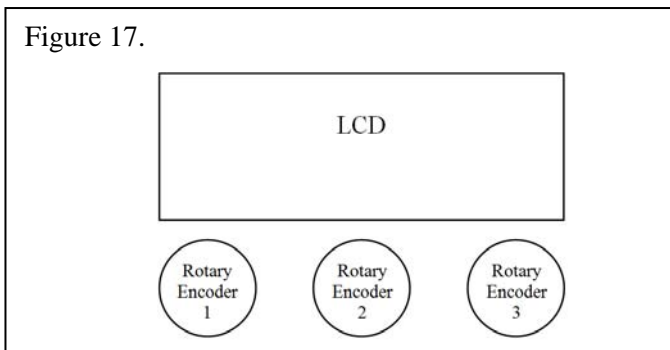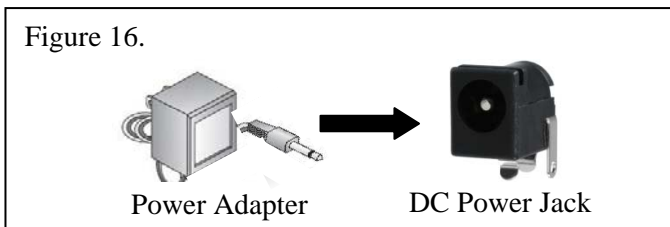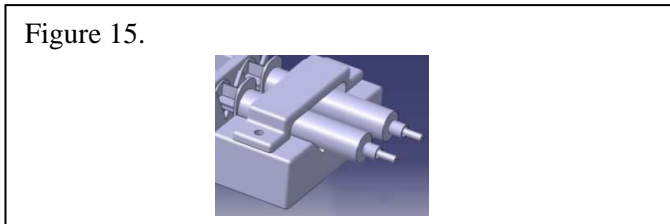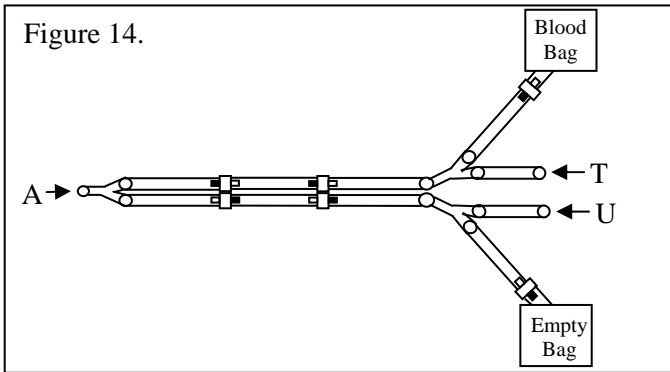
Figure 18.

Figure 19.

30. Rotate encoder 1 (refer to Figure 17) to the left until the plunger is fully retracted into the syringe. (Do not retract the plunger more than the syringe can handle.)

31. Press encoder 2 (refer to Figure 17) to set the position of the syringe when the plunger is fully retracted.

32. Rotate encoder 2 to adjust the volume flow rate (mL/min) in which the blood will be transfused into and out of the neonate.

33. Hold down encoder 2 for 3 seconds. Rotate encoder 2 to adjust the inner diameter of the syringe (measured in millimeters). Press encoder 2 to set the inner diameter of the syringe.

34. Rotate encoder 3 to adjust the total volume to be transfused into the neonate (measure in milliliters).

35. Prime the blood administration set and the umbilical catheter using standard hospital procedures.

    a. Squeeze and release the drip chamber until the solution/blood level has reached the Y connector connected to the umbilical catheter.

    b. Obtain two spare syringes of any size.

    c. Depress the plunger of each syringe to expel the air.

    d. Connect the spare syringes to ports T and U (refer to Figure 14 for the location of ports T and U).

    e. Pull out the syringe plunger until all of the air plus 8-10 mL of solution/blood are in the syringe.

    f. Remove the syringes connected to ports T and U (refer to Figure 14) and attach the ports T and U to the syringes currently in the syringe holder.

36. Follow hospital procedures to insert and secure the umbilical catheter in the vein.

37. Press encoder 3 to start the transfusion process. (It is recommended that medical personnel check on the neonate periodically to insure that the device is functioning correctly, as well as to monitor the vitals of the neonate.)

38. If the LCD screen does not reveal the correct input data after the start button is pressed, the press encoder 3 (refer to Figure 17) again to stop the transfusion process and input the variables again.

Figure 20.

Back Plate

Wing Nut

Figure 21.

Brace

Syringe Holder
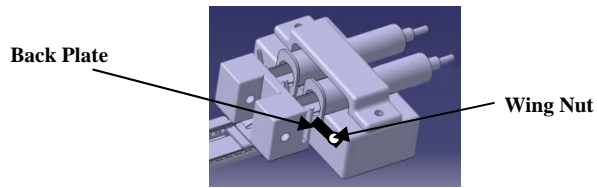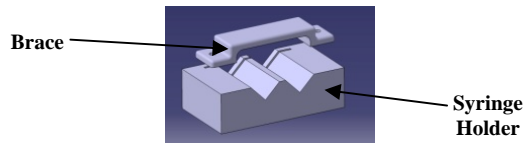
39. **ERROR ALERT**: If the system puts out an error message, then reset all of the input variables.
40. If the transfusion process needs to be paused mid-cycle, press encoder 1 (refer to Figure 17) to pause the cycle. To resume the cycle, press encoder 1 (refer to Figure 17) again and the cycle should begin where it last left off.
41. When the transfusion has been completed, a beeping noise will sound to alert the medical personnel.
42. When the transfusion process is completed, dispose of the tubing, syringes, Y connectors, and one-way valves. Also, unplug the power cord when the transfusion process is completed.

Appendix H: Safety Reports

Automated Exchange Transfusion Device Project 17
Dan Ciko, Brad Herron, Ashley May, Devon Stanforth

Safety Report 0.1 – Force Transducer Syringe Testing

1. EXECUTIVE SUMMARY
The planned experimentation for Thursday, March 12, 2009 is to test the force required to inject water out of several different size syringes. This experimentation is necessary to determine the capabilities the motor will need to have that we will be purchasing later. The experimentation will take place in Professor Gillespie's lab. Using a force transducer, we will measure how much force is required to push each different syringes' plunger in where the water flows out at a constant rate. The major risk involved with this experimentation will be containing the water while injecting it out of the syringe so it does not go everywhere and become a safety hazard. To avoid any harm to our eyes, we will use safety goggles while performing this experimentation.

2. EXPERIMENTATION PLANS
The experimentation will consist of pushing the force transducer against several different size syringes' plunger, while each syringes' plunger is fully extended out. The syringe will be held parallel to the ground, and the force transducer will be pushed against the syringe plunger while perpendicular to the ground at a constant rate to determine how much force is required to push the water out of the syringe. This experimentation will be performed with syringe sizes of 10, 20, and 25 mL, with two tests being conducted for each size syringe.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY
The material being used for this experimentation is the three different size syringes, the force transducer, and tap water.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS
No CAD drawings for this experimentation.

5. MANUFACTURING
There is no manufacturing necessary for this experimentation.

6. ASSEMBLY
There is no real assembly for this experimentation. The only requirement will be to fill the syringes with tap water, and then eject that water by pressing the force transducer against the syringe plunger.

7. DESIGN TESTING AND VALIDATION

The force transducer should give us a good idea of how much force is required to push the syringe plunger in to eject fluid from the syringe. This force will be the overlying force that the motor will need to overcome for the pumping mechanism to function properly. Once we have an idea of how much force is required, we will move on to purchasing a compatible motor.

Automated Exchange Transfusion Device Project 17

Dan Ciko, Brad Herron, Ashley May, Devon Stanforth

Safety Report 0.2 – Syringe Holder

1. EXECUTIVE SUMMARY

The planned experimentation for March 19, 2009 is to manufacture a holder for two syringes to be placed on. The material we plan to use to make this holder is the standard PVC plastic provided in the machine shop. We plan to use the mill and the band saw to manufacture this holder. We will not be using a CNC program for manufacturing as we will perform this particular experimentation by hand. The risks involved include possibly cutting ourselves, getting pinched fingers, and having debris fly into our eyes if the proper safety precautions are not followed. We will take the safety rules in the machine shop very seriously and handle the machine equipment very carefully to avoid any dangerous or harmful situations, and we will obviously wear safety goggles to avoid damage to our eyes.

2. EXPERIMENTATION PLANS PRIOR TO DESIGN COMPLETION

Measurements were taken on several dimensions of a 20 and 25 mL syringe to determine the syringe holder parameters prior to making the CAD model of the holder. Strength tests will be performed on this holder after manufacturing to determine if it is capable of withstanding the friction and other forces that will be put on it as the slider moves back and forth.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY

Because materials from the machine shop are both cost effective and conveniently available, we will use PVC plastic from the machine shop as the material for manufacturing this holder.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS

See attached CAD drawing.

5. MANUFACTURING

The manufacturing will be done in the machine shop (1302 GGB). We will first use the mill to cut away any unnecessary, excess material from the PVC plastic block we will be given, and then we will finish manufacturing the part with the band saw. Two parallel, triangular grooves will be cut into the top of the syringe holder to accommodate where the syringes will be placed. These grooves will be cut at a 45 degree angle for both sides of each groove, down to a depth of 1 inch from the top of the holder so that the syringes can rest easily in the grooves. Once the holder has been manufactured, we will briefly use sandpaper in the lab to smooth the edges out so there are no dangerous, sharp points.

6. ASSEMBLY
There will be no assembly of this syringe holder required. It will be manufactured in one piece. We will later assemble the holder to the slider outside of the machine shop to verify that the entire mechanism works properly.

7. DESIGN TESTING AND VALIDATION
We will test the holder and its strength along with its compatibility with the slider made in rapid prototyping in Professor Gillespie's lab. Once we verify that this sliding mechanism functions properly, we will test it with syringes in place. We will have our linear actuator move the slider back and forth eventually, but that will come later once further testing has been completed on the linear actuator.

Automated Exchange Transfusion Device Project 17

Dan Ciko, Brad Herron, Ashley May, Devon Stanforth

Safety Report 0.3 – Track Cutting

1. EXECUTIVE SUMMARY
The planned experimentation for Thursday, April 2, 2009 is to cut our ball bearing, stainless-steel track to a size that fits with the dimension limitations of our overall device. Cutting the track will take place in the GG Brown machine shop, and a band saw will be used to make the cuts. Only two, straight cuts will be necessary. The major risk involved with this procedure includes cutting ourselves with the band saw if the track is not properly held in place while cutting. To avoid any harm or injury, we will ensure that the track is safe and secure before running it through the band saw. We will also obviously wear safety goggles to avoid getting any debris from the track in our eyes.

2. EXPERIMENTATION PLANS
No experimentation will be performed on this track.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY
The track is a ball bearing track made of stainless steel.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS
See attached CAD drawing.

5. MANUFACTURING
Two cuts will be made on the track. One straight cut will be made across the slider at a distance of 6" from the back of the track. The other straight cut will be made across the slider at a distance of 6.5" from the front of the track. Again, the cuts will be made with a band saw.

6. ASSEMBLY
The track will be assembled later to the slider and the syringe holder, but that will take place at Iosix, Industries. No assembly will take place in the machine shop where the cutting will take place.

7. DESIGN TESTING AND VALIDATION
Once the track has been cut to size, we will validate that its dimensions are correct and it fits properly with our slider and syringe holder by assembling the three together. The track, after cutting, should still allow the smooth movement of our slider. This will allow the syringes to move back and forth, thus allowing the fluid to be ejected and injected properly.

Automated Exchange Transfusion Device Project 17
Dan Ciko, Brad Herron, Ashley May, Devon Stanforth

Safety Report 0.4 – Syringe Holding Plate

1. EXECUTIVE SUMMARY
The planned manufacturing for April 6, 2009 is to manufacture a syringe holding plate that will be placed on top of the syringes while the syringes lie in the grooves of the syringe holder. This plate will be bolted down to prevent movement or rotation of the syringes while the device is running, and it will be manufactured with an adjustable height so that various size syringes will work with our device. All components of the holder will be manufactured using the band saw. There will be two flat PVC plates, two L-shaped PVC plates, two pieces of compressible foam that will be put in between each flat PVC plate and each L-shaped PVC plate, and finally a rectangular plate made of plexi-glass to attach the two PVC plate sides together. The risks involved include possibly cutting ourselves with the band saw and having debris fly into our eyes if the proper safety precautions are not followed. We will take the safety rules in the machine shop very seriously and handle the machine equipment very carefully to avoid any dangerous or harmful situations, and we will wear safety goggles to avoid damage to our eyes. We will also handle the super glue very carefully when pasting all of the plate components together to make sure that we do not attach anything that should not be attached.

2. EXPERIMENTATION PLANS PRIOR TO DESIGN COMPLETION
Measurements were taken on the outside diameters of a 20 and 25 mL syringe to determine the height at which the syringes would be resting above the syringe holder to ensure that we made a plate tall enough to cover at least that height.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY
Because materials from the machine shop are both cost effective and conveniently available, we will be using PVC plastic and plexi-glass from the machine shop as the material for manufacturing this holder. The compressible foam to be placed in between the PVC plates to make the plate have an adjustable height was purchased from Home Depot. The top part of the top plate is being made of plexi-glass so the user can see through the glass to see where the syringe plunger position is at while the system is running.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS
See attached CAD drawing.

5. MANUFACTURING

The manufacturing will be done in the machine shop (1302 GGB). We will first use the band saw to cut two 1 in. by 1 in. flat plates of PVC. We will then cut two more equivalent plates of the same PVC, but these two plates will have an added height of PVC on one side to resemble an L shape. The added height will be approximately ½ in. We will then cut two 1 in. by 1 in. pieces of compressible foam to be placed in between each flat PVC plate and each L-shape PVC plate. The thickness of the compressible foam should be approximately ¼ in. We will then cut a piece of plexi-glass to lie across the two sides using the band saw. The dimensions for this plexi-glass piece are 1 in. by 4 in. Once all pieces have been cut using the band saw and attached with super glue accordingly, ¼ in. diameter holes will be drilled using the drill press through each side of the plate so that it will be able to be bolted down to the syringe holder properly.

6. ASSEMBLY

The assembly for this manufacturing consists of using super glue to attach all the plate components together, and then drilling holes in the plate and the syringe holder accordingly to fasten the plate to the syringe holder. Each hole in the syringe holder will have to be tapped once it has been drilled to allow for a bolt to be screwed in to secure the plate to the holder.

7. DESIGN TESTING AND VALIDATION

We will validate that this plate secures the syringes into the syringe holder and prevents the syringes from moving or rotating while the system is running by making sure that the syringes stay in place once the system has been turned on. We will also validate that both 20 mL syringes and 25 mL syringes can be locked into place using the plate to ensure that our device is compatible with multiple size syringes as required.

Automated Exchange Transfusion Device Project 17

Dan Ciko, Brad Herron, Ashley May, Devon Stanforth

Safety Report 0.5 – Mounting Board

1. EXECUTIVE SUMMARY

The planned manufacturing for April 9, 2009 is to manufacture a mounting board for all of our device components to be attached to. The board will be made from the wood boards available in the machine shop. The board will be cut into a rectangular shape with dimensions of 23.5 in. by 6.5 in. The cuts will be made using the band saw. Once the board has been cut to size, holes will be drilled into it in several places so that all of the device components can be attached to it such as the stainless steel track, the syringe holder, the linear actuator bracket, and the plexi-glass casing. The risks involved include possibly cutting ourselves while using the band saw, and having debris fly into our eyes if the proper safety precautions are not followed. We will take the safety rules in the machine shop very seriously and handle the machine equipment very carefully to avoid any dangerous or harmful situations, and we will obviously wear safety goggles to avoid damage to our eyes.

2. EXPERIMENTATION PLANS PRIOR TO DESIGN COMPLETION

Measurements were taken on several dimensions of all device components to be attached to the board to determine where each hole would need to be drilled into the board. There was no real experimentation done prior to this manufacturing other than measuring dimensions.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY

Because materials from the machine shop are both cost effective and conveniently available, we will use the wood boards from the machine shop to make our board out of, and we will use the bolts and screws available in the shop to attach the components onto the board.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS

See attached CAD drawing.

5. MANUFACTURING

The manufacturing will be done in the machine shop (1302 GGB). We will first use the band saw to cut the wood board to the correct size of 23.5 in. by 6.5 in. We will then measure on the board where each hole needs to be drilled to attach the device components onto the board. As stated above, the syringe holder, the stainless steel track, the linear actuator bracket, and the casing will all be attached to the board, so holes need to be drilled on several different parts of the board for these components to be attachable. The holes will be drilled using the drill press, and tapped using a manual tap.

6. ASSEMBLY
Once the board has been manufactured and all necessary holes have been drilled into it, we will assemble all of the components onto the board by screwing the bolts in where the holes were drilled and attaching the components.

7. DESIGN TESTING AND VALIDATION
We will validate that all components have been properly secured to the board by testing the functionality of the system later and making sure that there are no components moving that shouldn't be moving.

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Safety Report 0.6 – Casing

1. EXECUTIVE SUMMARY
The planned experimentation for April 13, 2009 is to manufacture a plexi-glass casing to be attached to the mounting board that will surround and protect our device. The casing will be made into four separate pieces using the laser cutter in the machine shop, and then the four pieces will be attached using superglue. We will also have a casing roof that will also be laser cut. This roof will be attached to one of the sides of the casing using hinges purchased at Home Depot. A CAD file of the four sides of the casing and the casing roof was made in CATIA, and that file was input into the laser cutting program so the laser cutter could properly cut the four walls of the casing and the roof. Once the four walls have been attached using super glue, we will drill four holes into the roof and four holes into one of the side walls of the casing to attach the hinges so the roof can open as a lid. The risks involved include possibly drilling into our fingers if we are not careful and having debris fly into our eyes if the proper safety precautions are not followed. We will take the safety rules in the machine shop very seriously and handle the machine equipment very carefully to avoid any dangerous or harmful situations, and we will obviously wear safety goggles to avoid damage to our eyes.

2. EXPERIMENTATION PLANS PRIOR TO DESIGN COMPLETION
Measurements were taken from where the holes were drilled in the mounting board to let us know how and where the casing would attach to the board. We also took measurements on the CAD file before laser cutting the walls so we knew where the hinges would need to be attached.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY
Because plexi-glass is the best material to use for the laser cutter in the machine shop, and because plexi-glass is transparent so people can still see the device running while enclosed in the casing, we chose to use that for our casing walls and roof. As stated above, the hinges used to attach the casing wall to the casing roof were purchased at Home Depot and are made of stainless steel.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS
See attached CAD drawing.

5. MANUFACTURING
We will not be doing much manufacturing in the machine shop. The laser cutter will cut all four casing walls and the casing roof for us. The only manufacturing we will have to

do is drilling holes into the casing wall and roof. These holes will be drilled using a power drill instead of a drill press because it is more convenient.

6. ASSEMBLY
As stated above, once the four walls have been successfully laser cut, we will attach all four of them together using super glue. We will then attach the hinges to the casing wall and roof to create a lid for the casing using shortened bolts that we will cut using a hack saw.

7. DESIGN TESTING AND VALIDATION
There will not be very much validation for the plexi-glass casing. As long as it stays upright and protects the device from outside debris, it will perform its function properly. We will validate that the lid opens properly and that the circuit board fits correctly into the corresponding side wall, but that will be the only validation required.

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Safety Report 0.7 – System Assembly

1. EXECUTIVE SUMMARY
The planned experimentation for Friday, April 3, 2009 is to assemble our device and all of its components. The plan is to first solder the circuit board materials onto the circuit board. Using a soldering iron presents a burning risk as the iron will be used at extremely high heats. To avoid any injury, we will use careful precision while soldering the circuit board together. To avoid any potential damage to our eyes, we will wear safety goggles throughout the entire procedure. Following the circuit board assembly, we will assemble the device with all of its components. Once the device has been fully assembled, we will test its functionality. Testing the functionality presents several risks that need to be considered before this experimentation can be conducted. The risks involved include being in contact with needles and the possibility of objects flying into the air if the linear actuator is run at a speed greater than it should be. Again, to avoid any injury, we will be extremely cautious while handling the needles and we will be wearing safety goggles while testing in case any objects fly into the air because of a system malfunction. All experimentation will take place at Iosix Industries in Ypsilanti, MI. To help eliminate the possibility of misusing any equipment, we will perform all experimentation with the supervision of our mentor Robert Vogt IV, CEO of Iosix Industries.

2. EXPERIMENTATION PLANS
For this particular experiment, we will focus on the functionality of a few key components of our device. After soldering the circuit board together and developing the program to run our system, we will first test the functionality of our linear actuator. The linear actuator will be powered by an AC power supply, and should move back and forth in a linear motion. We need to experiment with this linear actuator to make sure that it is capable of performing at extremely low speeds, which will be required for exchange transfusions. While this linear actuator is in motion, we will verify that the slider and slider holder are securely holding the syringes in place and that the syringes are allowed their necessary motion so fluid is able to flow in and out of them. We will also experiment with the one way valves that will go inside our tubing to ensure that they only allow fluid to flow in one direction as expected. This experimentation will consist of connecting the one way valves to the tubing and manually injecting water with a syringe through both sides of the tubing. If the one way valves function correctly, water should only actually flow through the tubing to the other side for one of the two ways. The proper functionality of the one way valves will eliminate the danger of blood traveling into places it shouldn't while the transfusion is taking place.

3. PURCHASED COMPONENT AND MATERIAL INVENTORY
There are several materials that will be assembled and used for our device. The electrical components consist of a circuit board, an AC power supply, and a linear actuator. The slider track is a ball-bearing track made from stainless steel, the actual slider is made from ABS plastic, and the syringe holder is made from PVC. The syringes are standard 20 or 25 mL syringes. The tubing is 2.5" diameter tubing. Other materials such as the one way valves and the y-junctions are made from plastic. Standard hospital IV bags and needles will also be used.

4. CAD DRAWINGS AND DESIGNSAFE SUMMARY FOR DESIGNED PARTS
See attached CAD drawings.

5. MANUFACTURING
The only manufacturing for this experiment will be the manufacturing/assembly of the circuit board. As stated earlier, this assembly will take place at Iosix Industries in Ypsilanti, MI with the help of Robert Vogt, IV, CEO of Iosix Industries. The circuit board will be assembled with all of its components using a soldering iron. There should be no manufacturing beyond this step. The rest of the experiment will be assembly and testing.

6. ASSEMBLY
The assembly is the main focus of this experiment. We need to ensure that our device can be assembled the way it was designed. First, we will connect our power supply to our linear actuator. We will then attach our linear actuator to the slider using a bolt. Next, we will secure the syringes into place in the slider holder by locking the syringes down with our syringe cover plate that rests above the syringes and is bolted down to the slider holder. We will then properly place the one way valves in the tubing, and then attach the tubing to the syringes and the IV bags. Finally, we will attach the y-junction to the two tubes coming from the syringes and insert a needle into the one IV bag that will be used to represent the baby.

7. DESIGN TESTING AND VALIDATION
Because water and blood have very similar fluid properties, we will experiment with water instead of blood. To help us visualize the path that the fluid will be taking while our system is in use, we will use red-colored water to represent the fresh blood and blue-colored water to represent the waste blood being removed from the baby. As stated earlier, we will use three IV bags. One will be used as the fresh blood bag, one will be used as the waste blood bag, and the last one will have a needle going into it to represent the baby.

We will validate that our user input function works correctly by entering in different values into the LCD screen for volume flow rate, total volume to be transfused, and syringe inner diameter. We will manually measure these values for the first few experiments to ensure that the numbers displayed on the screen are correct. We will then validate the functionality of our linear actuator, our slider mechanism, and our one way

valves by watching the system run through a few tests, making sure along the way that all fluid is flowing properly into its intended destinations.

Appendix I: Prototype Photo