Faculdade de Ciências e Tecnologia
Universidade de Coimbra
Departamento de Física

Intelligent Sensing Anywhere

EasyBreathing@Home

Remote Vital Signals Monitoring in Respiratory Failure

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“*O Homem é do tamanho do seu sonho.*”

Fernando Pessoa

Resumo

A Doença Pulmonar Obstrutiva Crónica é considerada pela Organização Mundial de Saúde um problema de saúde pública, afectando cerca de 270 milhões de pessoas em todo o Mundo. Em Portugal, o Observatório Nacional de Doenças Respiratórias (ONDR) estima que cerca de 540 mil portugueses sofram desta patologia, número que se encontra porém, subestimado, com muitos casos mal diagnosticados e outros ainda por diagnosticar.

A utilização da Oxigenoterapia Domiciliária de Longa Duração possibilita aos doentes reduzirem as consequências desta patologia, permitindo aliviar a pressão na artéria pulmonar e reduzir o esforço cardíaco. Neste tipo de tratamento, o oxigénio é utilizado por um período de tempo igual ou superior a 15 horas diárias, o que permite ao doente aumentar o seu nível da actividade física e melhorar substancialmente a sua qualidade de vida. Assim, estima-se que cerca de 2 milhões de pessoas em todo o mundo beneficiem deste tipo de terapia.

Actualmente, nos sistemas disponíveis no mercado, o controlo do débito de oxigénio é feito manualmente, o que leva, por um lado, a que o mesmo não coincida com as necessidades reais e dinâmicas do paciente, assim como a desperdícios de oxigénio, aumentando os custos associados à terapia.

A utilização de um sistema de controlo do débito em tempo real com base em leituras contínuas da saturação sanguínea de oxigénio permite, não só estabilizar de modo mais rápido e eficaz o doente, como também reduzir custos associados ao desperdício de oxigénio, assim como os gastos com o internamento hospitalar, provocados pelo estado continuado de dessaturação e as suas consequências negativas. Por outro lado, tendo em conta que a necessidade de oxigénio varia consoante o grau de actividade física do momento, os dados relativos ao mesmo são também relevantes no cálculo do volume de oxigénio a fornecer.

Esta tese fornece uma descrição da inclusão de uma válvula proporcional electrónica e respectivo *hardware* de actuação num sistema integrado, bem como do controlo dinâmico da abertura de saída, através de um microcontrolador, tendo por base dados de oximetria de pulso obtidos em tempo real. É igualmente apresentado um pequeno estudo realizado com o objectivo de implementar um controlador PID no dispositivo.

Abstract

Chronic Obstructive Pulmonary Disease (COPD) is considered by the World Health Organization a public health problem, affecting at least, 270 million people around the world. In Portugal, the ONDR estimates that more than 540 thousand people suffer from that pathology, a number which is certainly underestimated, considering the large number of unknown and under diagnosed cases.

The use of Long-Term Oxygen Therapy (LTOT) at home enables the patients to moderate the consequences of this condition, reducing the pressure in the pulmonary artery and the cardiac effort. In this type of treatment, the oxygen is used at least 15 hours per day, allowing the patients to increase their level of physical activity and substantially improve their life's quality. Thus, it is estimated that about 2 million people worldwide benefit from such therapy.

Currently, in the systems available in the market, the control of the oxygen's flow is done manually, which leads on the one hand, that it doesn't match with the real and dynamic needs of the patient, and to the waste of oxygen, increasing the total costs of the therapy.

The use of a flow control system in real time using continuous readings of blood oxygen saturation can, not only stabilize quickly and effectively the patient, but also reduce the costs associated with the waste of oxygen, as well as the internment's spending in the hospital, caused by the continued state of desaturation and its negative consequences. Moreover, given that the need for oxygen varies with the degree of physical activity, the information on it is also relevant in the calculation of the oxygen's volume to be supplied.

This thesis provides a description of the inclusion of an electronic proportional valve and its hardware of actuation in an integrated system, as well as the dynamic control of the valve's opening by means of a microcontroller, based on data from pulse oximetry obtained in real time. It also provides a brief description of a study done to implement a PID controller in the system.

Acknowledgments

First, I would like to thank to the project's coordinators Professor Carlos Correia and Professor José Basílio Simões for making this project possible.

I also would like to thank to Engineer José Eduardo Faria for his guidance in the first stage of the project and also to Engineer Nuno Varelas for his support.

I'm grateful to Engineer José Luís Malaquias for his valuable help and for the innovative ideas that contributed to the project enrichment and also to Engineer Catarina Pereira for her assistance in the project improvement.

I'm also thankful to Engineer Soraia Rocha, her advices, sympathy and guidance along the course of the project were crucial to its development.

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To my parents and to my sister Marisa, for always being there for me; without them, nothing of this would be possible.

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To all of them, me special thanks.

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Acronyms and Definitions

CEI	*Centro de Electrónica e Instrumentação*
CHC	*Centro Hospitalar de Coimbra*
COPD	Chronic Obstructive Pulmonary Disease
CPU	Central Processing Unit
DODS	Demand Oxygen Delivery Systems
FiO_2	Fraction of Inspired Oxygen
IDE	Integrated Development Environment
IEEE	Institute of Electrical and Electronics Engineers
INE	*Instituto Nacional de Estatística*
ISA	*Intelligent Sensing Anywhere*
LED	Light Emitting Diode
LOX	Liquid Oxygen
Lpm	Liters per minute
LTOT	Long-Term Oxygen Therapy
NOTT	Nocturnal Oxygen Therapy Trial
OEM	Original Equipment Manufacturer
ONDR	*Observatório Nacional das Doenças Respiratórias*
$PaCO_2$	Partial Pressure of Carbon Dioxide
PaO_2	Partial Pressure of Oxygen
PC	Personal Computer
PIC	Programmable Interface Controller
PID	Proportional, Integral, Derivative
PWM	Pulse Width Modulation
RS-232	Recommended Standard 232
RS-485	Recommended Standard 485
SaO_2	Arterial Oxygen Saturation
SCI	Serial Communication Interface
SpO_2	Saturation of Peripheral Oxygen
TOC	Transtracheal Oxygen Catheters
TTL	Transistor to Transistor Logic
TxD	Transmitted Data
UART	Universal Asynchronous Receiver Transmitter
USB	Universal Serial Bus

1. Introduction

1.1. Motivation

Chronic Obstructive Pulmonary Disease (COPD) is an incurable and slow progress disease that results in air obstruction and breathing difficulties. In Portugal, more than 540 thousand people suffer from COPD, a number which is certainly higher, considering the large number of unknown and under diagnosed cases. Besides, the situation becomes worst in elderly people, since COPD frequency increases substantially with age. In fact, the Portuguese ONDR estimates that, above 60 years of age, the prevalence of the disease is 138 cases per 1.000 inhabitants in men and 62 cases per 1.000 inhabitants in women. Furthermore, this disease is the sixth leading cause of death worldwide; the fifth in Europe and other developed countries, while in the U.S.A. is the fourth.

Since this is a chronic disease, the patient lives with the illness, on average 20 years, thus representing a burden for patients and for the state. In Portugal, the supplemental oxygen used in their treatment is fully reimbursed by the national health system. The treatments available to COPD relieve the pulmonary damage and slow the disease progression, consisting in the daily supplying with additional oxygen; this therapy is well known as Long Term Oxygen Therapy (LTOT). This treatment uses a constant and previously prescribed flow of oxygen, delivered from a gas cylinder, a liquid oxygen system or an oxygen concentrator for more than 15 daily hours. [1] However, oxygen levels vary with activity intensity and also with environment conditions, leading the patients both to hypoxia and oxygen's wastage periods. Several studies demonstrated that, over the course of the patient's daily activities, the arterial oxygen saturation falls below an acceptable threshold (SpO_2<90%). However, excessive rates of oxygen also have negative consequences; in patients with hypercapnia, too much oxygen can depress ventilatory drive thereby aggravation of CO_2 retention; increased levels of CO_2 in the bloodstream can lead to acidosis. This indicates that patients aren't realizing the full potential benefits of the oxygen therapy, because their oxygen's demand isn't well matched with the delivered flow.

Additionally, we are in the era of the "*Medicine of the Movement*." The movement, especially walking on foot, is highly recommended to respiratory failure patients, since it is important, not only to the respiratory mechanics, but also to the improvement of respiratory physiology. However, the leg muscles are big consumers of oxygen and lead to large decreases in oxygen saturation, which creates discomfort. As

well, the physical activity is directly linked with the consumption of oxygen. Thus, for low levels of physical activity, the output flow needed is lower; while for higher levels of physical activity are required higher flows. So, this information is certainly relevant in the output flow determination.

Furthermore, the oxygen delivery is controlled manually, but, special in acute ill patients, the achievement of safe and stable levels of oxygen saturation is not only difficult, but also time consuming, once it needs constant attention of the medical personnel.

Previous studies attended in the automatic control of the supplemental oxygen deliver and demonstrated its potential in maintaining optimal blood oxygen saturation. The automatic control showed, therefore, to be more efficient in the patient management, especially in the systems that applied the pulse oximeter readings as feedback signal to a controller. [2]

From an economic standpoint, the automatic control of the oxygen deliver generates the interest of multinationals that supply oxygen systems, which enables not only a significant saving of oxygen, but also resources' savings. These companies have technicians who travel several times to the patients' home to exchange the oxygen bottles; as this exchange is done in fixed periods of time, sometimes, are made unnecessary travel. Moreover, this area is also poorly explored.

1.2. Objectives

The aim of this project is to build a system to automatically and efficiently control the deliver of oxygen to a chronic ill patient, by controlling the flow based on data collected in real time. As well, the proposed device intends to be a homecare solution to be used in home oxygen therapy. With a well known way of non-invasive monitoring technique- the pulse oximetry, it's possible to access the oxygen saturation, observing the patient's needs and also monitoring the treatment's efficiency. By applying a 3-axis accelerometer in the system is possible to monitor the patient's movement and therefore it's possible to control the oxygen flow by the movement. That represents a great advance for chronic ill patients that can have a dynamic therapy and consequently, they can improve their quality of life. With the projected system, the oxygen flow delivered from the source to the patient can be controlled dynamically and not statically, with a fix flow of oxygen, which is the major problem in the current available systems. Additionally, with the dynamic therapy it's expectable to avoid the majority of the dessaturation periods and the discomfort caused by them.

1.3. Document Structure and Organization

This document is composed by 9 chapters; each one corresponding to different contends and blocks of work.

This chapter is an introduction, with the project's motivation and objectives.

Chapter 2 describes the project management; its team members and the supervising at the different institutions that cooperated with this work. The chapter also refers the project planning.

Chapter 3 is related to the theoretical background; it contains the mainly information achieved before the beginning of the system's development.

Chapter 4 is referred to the State of the Art; it's a brief description of the projects in course that have the same goals as ours and also the works that use identical concepts.

Chapter 5 describes the system development, providing its logic architecture and the project requirements and features. It also refers the miniature proportional valve and the driver board's integration in the system and its actuation by the microcontroller, as well as the algorithm developed.

Chapter 6 relates the effort done in order to implement a PID controller in the system. It refers not only the analytic development, but also the study in hospital's ambience, where was collected data in the patients, which was latter analysed to calculate the PID parameters.

Chapter 7 explains the prototype's tests as well as their results; first in the laboratory and latter at the hospital, using medical oxygen and real saturation readings from the patients.

Chapter 8 is a brief description of the future works to be developed to increase the functionality of the system.

Chapter 9 is the conclusion, with a brief appreciation of the prototype obtained and its features.

2. Project Management

2.1. Project Team Members

The project team was composed by two students of the integrated master degree in Biomedical Engineering from the Faculty of Sciences and Technology, University of Coimbra and the project coordinators from CEI (*Centro de Electrónica e Instrumentação*) of the Physics Department from the Faculty of Sciences and Technology of the University of Coimbra. It also counted with the technical supervising at ISA- *Intelligent Sensing Anywhere*, and at the Pneumology Service in CHC (*Centro Hospitalar de Coimbra*). The elements that composed the team work are presented in the next table.

Name	Contribute	E-mail Contact
Patrícia Borges	Student	pborges@isa.pt
Neuza Aguiar	Student	naguiar@isa.pt
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José Luís Malaquias	Supervisor	jmalaquias@isa.pt
Joaquim Moita	Supervisor	joaquimmoita@chc.min-saude.pt
José Eduardo Faria	Supervisor	jfaria@isa.pt

Table 1: Project team members, the students, the University coordinators, the ISA technical supervisors and the supervisor at CHC.

2.2. Project Supervising

The project supervising was divided between CEI, ISA and the Pneumology Service from CHC.

2.2.1. Supervising at CEI

The Electronics and Instrumentation Centre of the University of Coimbra develops work and research in different areas, such as nuclear, atomic and biomedical instrumentation, optical signal processing, instrumentation for plasma physics and telemetry and industrial control. The centre includes not only researchers and schoolmasters, but also technicians and students. The supervising at CEI consisted in a few informal meetings where was given to the student some documentation and also clarified some questions.

2.2.2. Supervising at ISA

ISA, *Intelligent Sensing Anywhere*, was founded in 1990 by a group of engineers from the University of Coimbra, as a spin-off company. This company develops solutions in different areas, as remote management for utilities, including telemetry and remote metering solutions to different kinds of fuel, electricity and water, machine-to-machine solutions and, more recently, healthcare and medical solutions.

At ISA, the work supervising was done in the first stage by Engineer José Eduardo Faria, as the technical supervisor. In the second stage, the supervision was conduced by Engineer Soraia Rocha, who became the project's technical supervisor. During the supervising at ISA, the project also had the cooperation of Engineers José Luís Malaquias and Nuno Varelas.

The supervising at ISA consisted of periodic meetings at ISA's installations where were discussed not only the project evolution, but also the tasks and problems that the students were facing. Besides, the students always kept in contact with their technical supervisor to clarify doubts and questions that arose during the project.

2.2.3. Supervising at Pneumology Service, CHC

The supervising at the Pneumology Service from CHC consisted in several meetings with Doctor Joaquim Moita at CHC's installations where were discussed some medical considerations that must be well thought-out during the building of a medical system. The cooperation of this service and particularly from Doctor Moita enabled the prototype's tests in hospital's ambience, as well as patient's data collecting in some blocks of work. His guidance allowed the establishing of several features of the system and also the basis to the flow's control algorithm.

2.3. Project Planning

The project was mainly divided in two functional areas, which were the pulse oximeter integration, its data acquisition and process, and the miniature proportional valve and the driver board's integration and actuation by means of a microcontroller.

Neuza Aguiar was responsible for the firmware development to the pulse oximeter integration, while Patrícia Borges was responsible for the firmware development to the proportional valve's integration and actuation. The students cooperated to develop a global controller algorithm, where the oximeter data is received and according to it, the microcontroller actuates the valve.

In the first semester both of the students worked together in many tasks, especially in theoretical background area and in the system architecture, features and specifications definition. The students have also done some tasks, as the posters to the project's presentations and also two presentations to *ISA Intellicare* staff, about the project and about the PIC microcontrollers programming. In the project's first stage, the student was also focused in the system components, consequently there were made many contacts with companies that provide pulse oximeters and miniature proportional valves.

In the second semester, there were a division of work between Patrícia and Neuza. Each student followed a plan of tasks given by the project technical supervisor, which was focused in each functional area. Furthermore, the student has also done a few studies in order to implement a PID controller in the system, including the data collecting at the Urgency Service of CHC.

Following the algorithm's development by each student, they built a global algorithm to control the integrated system and they have done some computer tests. With those tests, they proceeded with the optimizations in each algorithm's part.

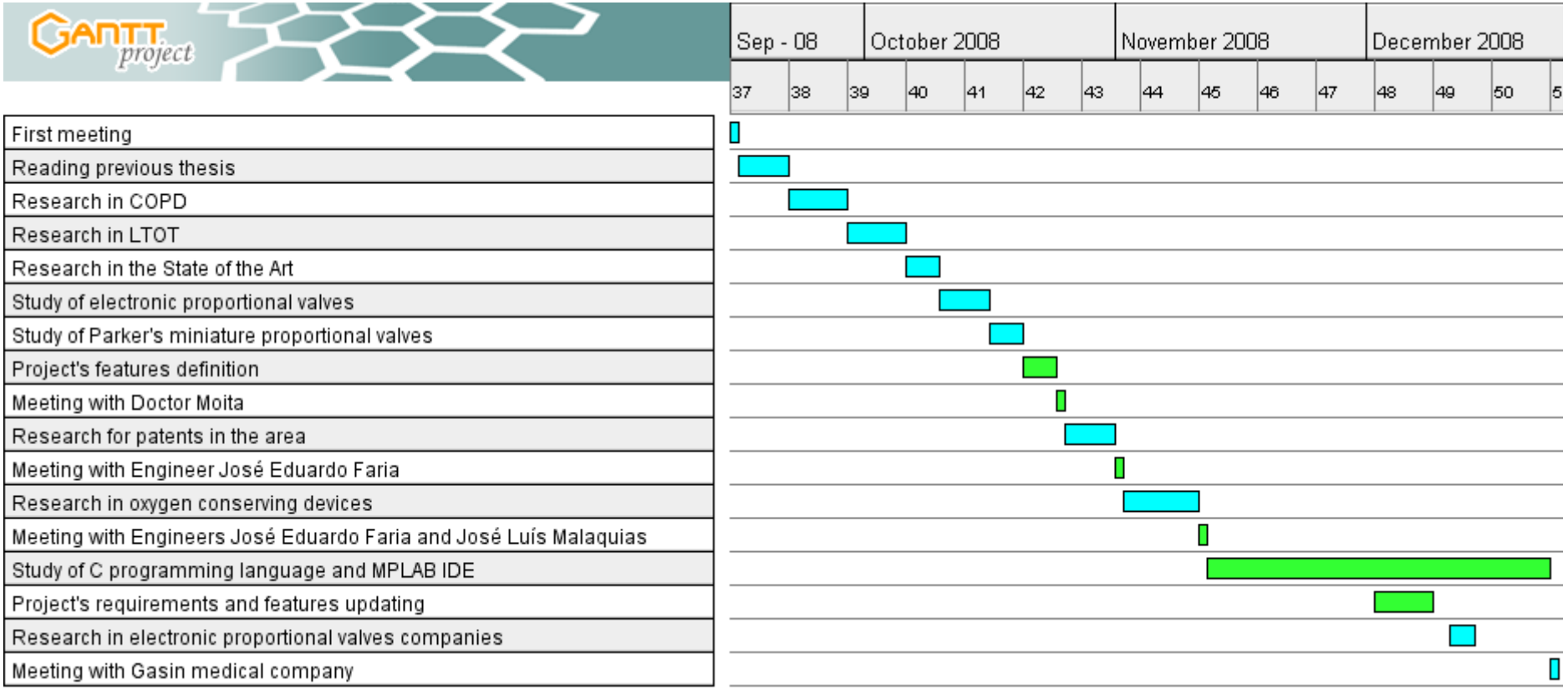
The last stage of the project was done together again and consisted in the prototype's experiments that were done at the Pneumology Service. Those tests enabled the analysis of the system's behaviour in real conditions, since they were done in a patient and with medical oxygen.

2.4. *Gantt* Charts

There are shown next the project's *Gantt* charts with the mainly tasks of the project, as well as their planning and the resources used. The colours' scheme applied to represent the resources used in the tasks is the follow:

Task done by Patrícia Borges

Task done by both students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First meeting
Reading previous thesis
Research in COPD
Research in LTOT
Research in the State of the Art
Study of electronic proportional valves
Study of Parker's miniature proportional valves
Project's features definition
Meeting with Doctor Moita
Research for patents in the area
Meeting with Engineer José Eduardo Faria
Research in oxygen conserving devices
Meeting with Engineers José Eduardo Faria and José Luís Malaquias
Study of C programming language and MPLAB IDE
Project's requirements and features updating
Research in electronic proportional valves companies
Meeting with Gasin medical company

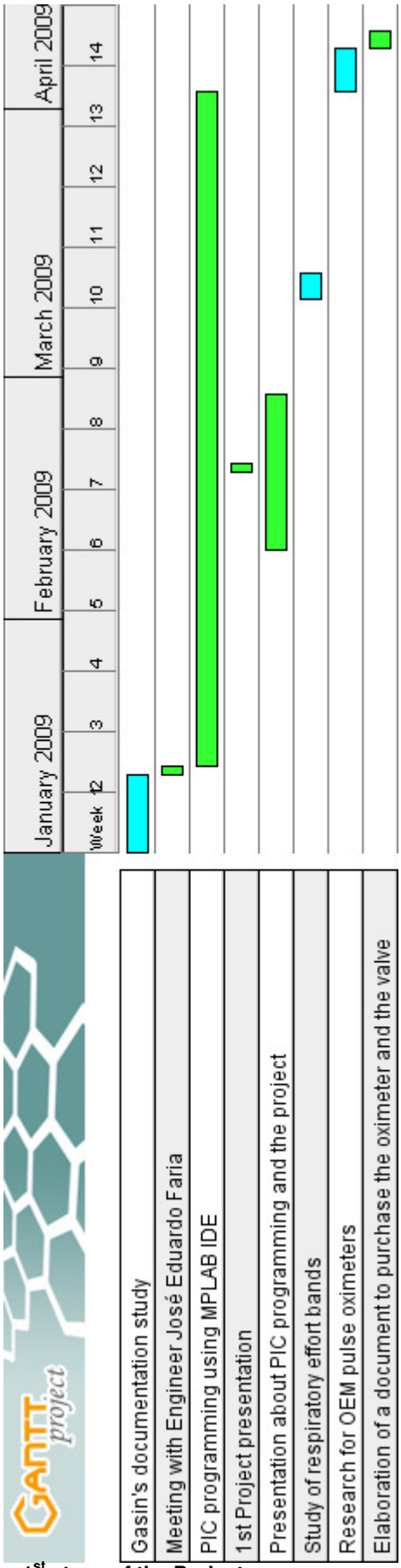


Figure 1: Gantt charts from the 1st stage of the Project.

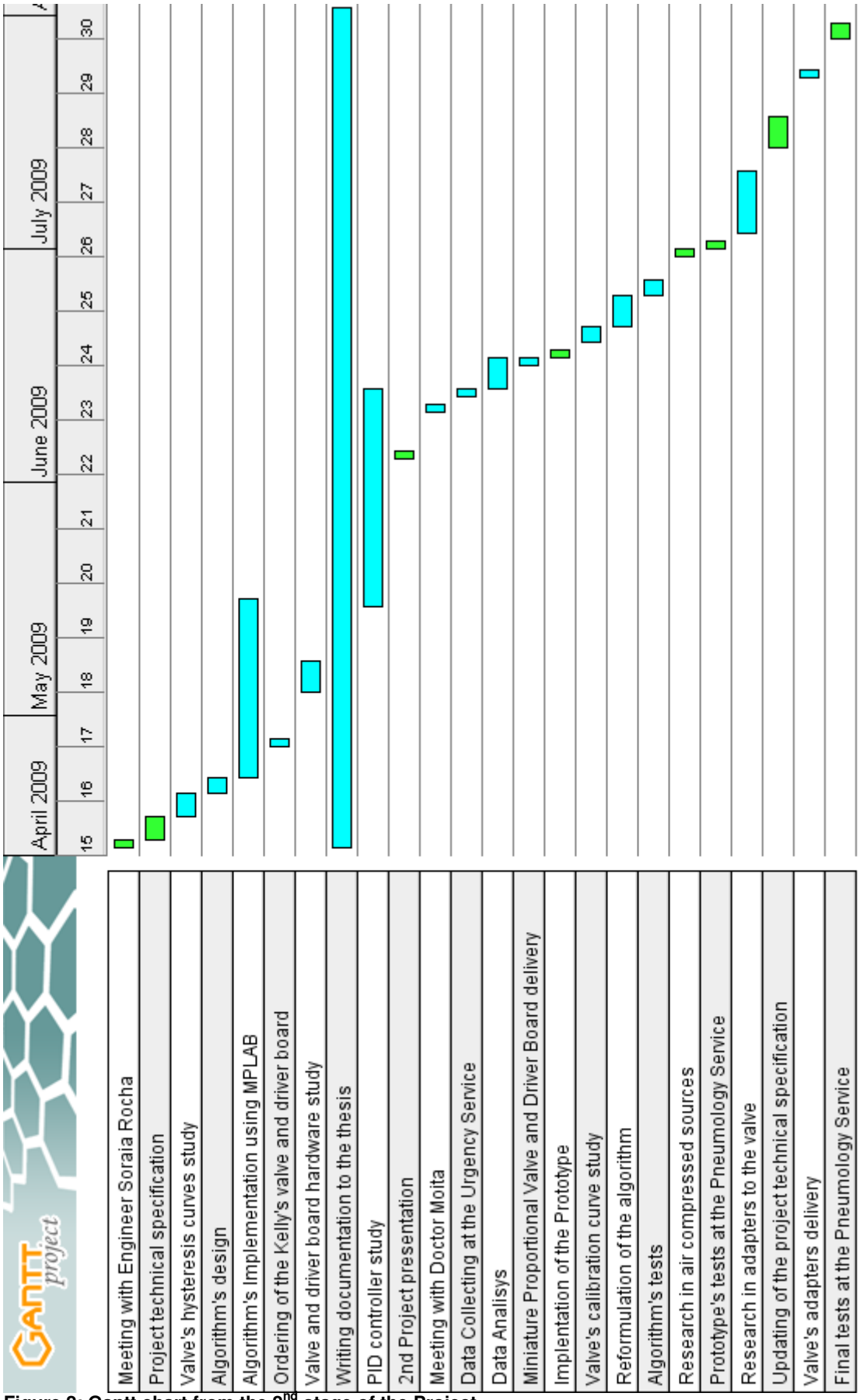


Figure 2: Gantt chart from the 2nd stage of the Project.

3. Theoretical Background

3.1. The Respiratory System

The respiratory system is composed by two parts, the upper respiratory system and the lower respiratory system. The upper part is composed by the nose, nasal cavity, paranasal sinuses and pharynx. These passageways are responsible for the filtration, warming and humidification of the air, protecting the surfaces of the lower respiratory system. On the other hand, the lower respiratory system consists of the larynx, trachea, bronchi, bronchioles and alveoli of the lungs, where the gas exchanges occur. [3]

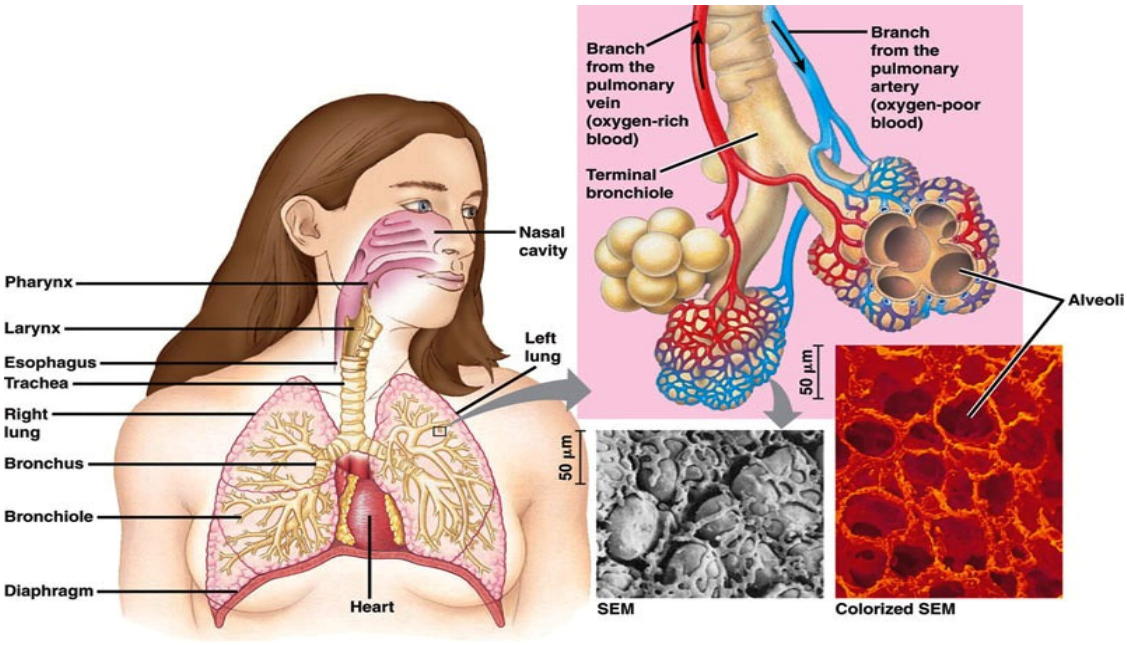


Figure 3: Diagram of the **Respiratory System**, showing its components and a diagram of the alveoli (bronchiole's terminals) with the capillaries to the gas exchanges. [4]

This system is responsible for five major functions: providing gas exchanges between air and circulating blood; moving air to and from the respiratory surfaces of the lungs; protecting respiratory surfaces from dehydration, temperature changes and environmental variations; production of sounds used in oral speaking and providing olfactory sensations to the central nervous system from the olfactory epithelium in the superior part of the nasal cavity.

The respiratory system is controlled by the brain's respiratory centres, which are coordinated with changes in cardiovascular system, fluctuations in blood pressure and cardiac output. The cerebral centres are responsible for the regulation of the respiratory muscles' activity, such as the diaphragm and the external intercostal muscles, by the generation of impulses, and they also control the respiratory minute

volume by adjusting the frequency and depth of pulmonary ventilation. The motor neurons in the spinal cord, which are generally actuated by respiratory reflexes, can also control the respiratory muscles. [4]

3.2. Respiratory Failure

Respiratory failure is characterized by the inability of the lungs to adequately exchange oxygen and carbon dioxide between the environment and the body tissues, transferring oxygen from inhaled air into the blood and carbon dioxide from the blood into exhaled air. [5] The direct consequence of this situation is the inability to keep the arterial blood gases at normal level, leading to $PaCO_2$>49 mmHg and/or PaO_2<60 mmHg, in room air (0.21 FiO_2). [6]

The basis of respiratory failure may be at several points; it may be a failure of the exchange of oxygen and carbon dioxide within the alveoli in the lungs; a failure of respiratory muscles to expand the lungs; or a failure in brain centres controlling respiration. [5]

The respiratory failure can be either chronic or acute and it's classified as hypoxemic or hypercapnic. In hypoxemic respiratory failure there are low levels of PaO_2, which is lesser than 60 mmHg, while the $PaCO_2$ is at normal or low levels. In hypercapnic respiratory failure the $PaCO_2$ is greater than 50 mmHg and hypoxemia is also common on hypercapnic patients while they're breathing ambient air. Chronic respiratory failure is developed over several days or even longer time periods, while acute respiratory failure is developed during periods ranging from minutes to hours. [7]

Chronic Obstructive Pulmonary Disease (COPD) is a term used to describe two related lung diseases: Chronic Bronchitis and Pulmonary Emphysema. Chronic bronchitis is an inflammation and eventual scarring of the bronchi, while Emphysema is an enlargement and destruction of the alveoli within the lungs. Patients with COPD may suffer from both of these problems, leading them to breathing difficulty because of the smaller air passageways, which are also congested with mucus and also to partially destroyed alveoli. The most important risk factor, which is also COPD cause, is smoking, in fact, about 80 to 90% of COPD cases are caused by smoking and a smoker is much more probable than a non-smoker to suffer/die of COPD. [8]

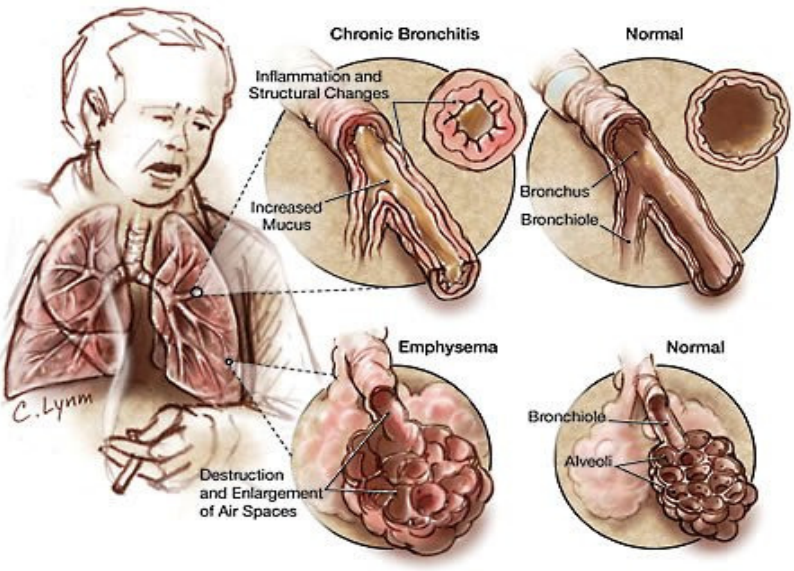


Figure 4: Chronic Bronchitis and Pulmonary Emphysema: comparison between healthy bronchi, bronchiole and alveoli with Chronic Bronchitis and Pulmonary Emphysema ones. [8]

3.3. Oxygen Therapy

The goal of therapy in respiratory failure is to achieve and maintain adequate gas exchanges. [6] The oxygen therapy consists of oxygen's administration in higher concentrations than in ambient air to reduce the effects caused by the oxygen's lack. [9] Furthermore, the treatment with supplemental oxygen is commonly used to improve quality of life, reliving hypoxemia, which raises the alveolar tension, reduces the breathing's work and decreases the myocardium's effort. [6] Thus leads to an increasing of exercise tolerance, energy and level of activity, which is fundamental to keep healthy. [9]

Treatment of patients with oxygen at home is called either *Home Oxygen Therapy* or *Domiciliary Oxygen*. [10] Oxygen Therapy can be prescribed in two different ways: short and long term. Short term oxygen therapy is applied when the patient is recovering from acute lung disorders, like pneumonia, that causes low levels of oxygen. On the other hand, in long term oxygen therapy (LTOT) the patient receives at least 15 hours per day of oxygen to achieve the maximum benefits of the treatment. [9] LTOT has several benefits, such as increases survival, reduces pulmonary artery pressure, alleviates right heart failure, neuropsychological improvement, and also the improvement of sleep quality and the reduction in cardiac arrhythmias. [11]

Non-continuous oxygen therapy is usually prescribed to alleviate or diminish pulmonary symptoms in particularly situations, such as the sleeping hours. Nocturnal oxygen is only applied during sleep due to the desaturations that occur in this period. [10]

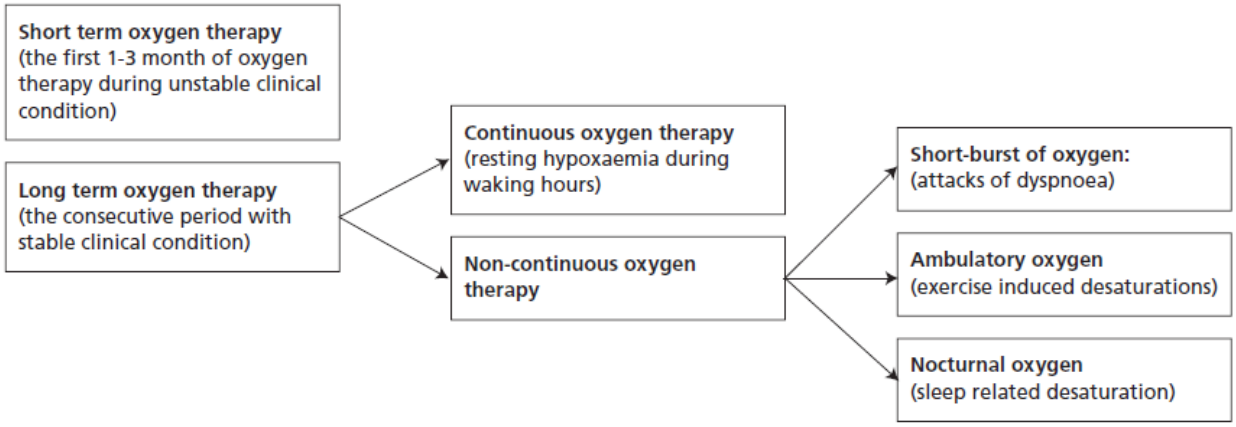


Figure 5: Terminology of **Home Oxygen Therapy**. [10]

3.4. Oxygen delivery systems

There are three oxygen sources available for LTOT at home: oxygen concentrators, liquid oxygen systems (LOX) and high-pressure cylinders, filled with gaseous oxygen. [12] The oxygen is administrated through oronasal devices, such as nasal cannulae and catheters and different kinds of face masks, like Venturi mask. [6]

3.4.1. Oxygen Concentrators

Air is mainly composed by nitrogen, about 78%, oxygen at 21% and other gases, such as argon and carbon dioxide, in a small amount, 1%. [13]

An Oxygen Concentrator is an electric device that separates oxygen from other gases in atmospheric air, through the employment of a molecular sieve. [14] Thus, allows the continuous delivering of concentrated oxygen to the patient and takes away the obligation of changing the oxygen's cylinders.

The atmospheric air is drawn into the unit through a series of particle filters; it's compressed, into a compressor, and then passes through a heat exchanger to dissipate the heat of compression. [12] Then, it passes through a cylindrical molecular sieve bed, the "zeolite tower", filled with a zeolite material, *aluminium silicate*. The nitrogen molecules are captured and selectively adsorbed onto the zeolite granules, a reversible reaction. When the zeolite granules become saturated with nitrogen, the system changes to a second tower through a switch valve, to achieve a constant production of oxygen. In the meanwhile, the first tower is vented off to ambient atmospheric pressure and the adsorbed nitrogen is released from the granules, which become capable of adsorbing nitrogen again. [15] The system stores the concentrated oxygen in a small cylinder for delivery to a flow meter. [12]

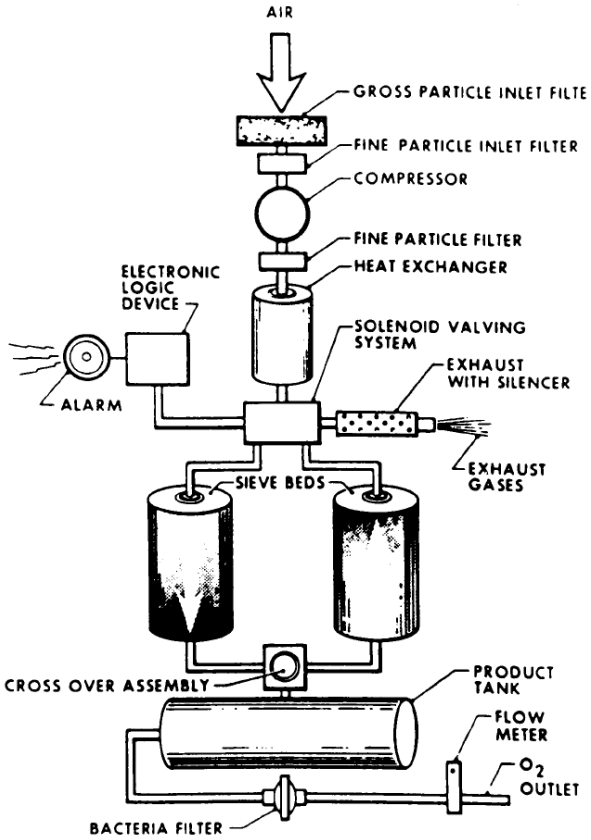


Figure 6: Diagram representing a **molecular sieve oxygen concentrator**. [12]

With an oxygen concentrator, the oxygen is delivered in high concentrations, although they don't get a purity of 100%. The Fraction of Inspired Oxygen (FiO_2) will depend on the flow rate; with lower flow rates the oxygen's concentrations are higher. [12] The produced oxygen's purity is presented in the next table.

Flow Rate	Oxygen's Concentration
≤1 L/min	$FiO_2 \leq 0.97$
1-2 L/min	$FiO_2 \geq 0.95$
3-5 L/min	$FiO_2 \geq 0.90$
>5 L/min	$FiO_2 < 0.90$

Table 2: The Output of oxygen concentrators, the correlation between the oxygen flow rate and the oxygen's purity (FiO_2). [12], [14]

3.4.2. Liquid Oxygen Systems

Liquid Oxygen (LOX) Systems store large amounts of oxygen in small and compact containers at low pressure, as in the liquid form 1L of LOX is equal to 840 L of gaseous oxygen. [12] The system is usually composed by two units: a stationary reservoir and a lightweight portable unit. When the patient is at home uses the

stationary source as the primary source and, when leaves home, fills the smaller portable unit. [16] Both the stationary and the portable units are constructed as a thermos bottle, with an inner and outer container with vacuum between them, keeping the oxygen at -297.3ºF and under a pressure of approximately 20 psi. [12], [17]

Figure 7: LOX containers, showing several stationary units and also a portable one. [18]

In the lower portion of the container, the oxygen is in the liquid form, while in the upper portion it's in the gaseous form, due to the evaporation. The flow control valve deliveries oxygen and it's opened when the unit it's being used by the patient. This creates a pressure gradient between the oxygen in the upper portion of the container, which is the head pressure, and the atmospheric pressure. LOX them passes through the vaporizing and warming coils where it is converted into gas available to the patient's use. When the head pressure falls below a certain limit, LOX is collected from the container's base to guarantee a constant flow to the patient. The LOX is under pressure and the ambience temperature causes evaporation of the liquid to gas, creating more pressure in the reservoir. In a certain pressure point, a primary relief valve opens to vent to the outside.

The container also has a secondary relief valve that opens when the pressure reaches 10 psig above the working pressure, if the primary valve fails. [17] The quick connect attachment is the connection to fill the portable unit. [12]

The major advantage of LOX systems is not only the large quantity stored, but also its purity; it contains 100% oxygen in a small and even portable unit. However, LOX systems are very expensive, since they have two reservoirs and the stationary reservoir must be filled by a technician every 10-14 days, which raises the costs. In addition, the transfilling of LOX from the stationary reservoir to the portable unit is complex and can cause thermal burns in the skin due to the low temperatures, if the patient touches the LOX. [12]

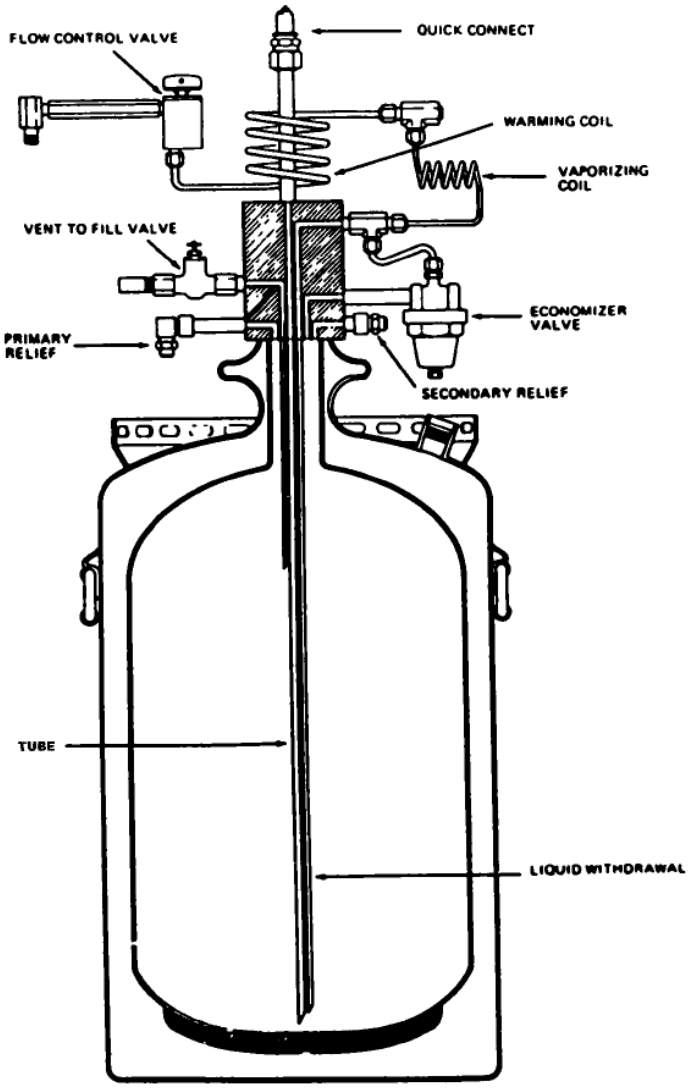


Figure 8: Components of a **stationary LOX container**. [12]

3.4.3. Oxygen Cylinders

Oxygen cylinders are made of steel or aluminium and they're filled with compressed gaseous oxygen under high pressure. [19] They're available in a wide variety of sizes, ranging from large stationary to small portable ones. [9] Each company has developed alternative sizes of small portable cylinders. Usually, the portable cylinders, also called ambulatory oxygen, are made in aluminium which can have less 50% of the weight when compared with similar ones in steel. [12]

The cylinder's dimension is identified alphabetically from A, which is the smallest one, to E, which is the bigger one. H and G cylinders are fixed and the others are portable. The major limitation of this oxygen delivery system is the limitation of the gas volume. [12] In the next table is presented the duration of the each cylinder at 2 L/min of flow:

Size	Litres (L)	Time
H	6908	57h, 33 min
G	5302	44h, 11 min
M	3625	30h, 12 min
E	616	5h, 8 min
D	352	2h, 56 min
B	150	1h, 15 min
A	76	38 min

Table 3: Oxygen Cylinders size, storing capacity and time duration (at 2L/min). [12]

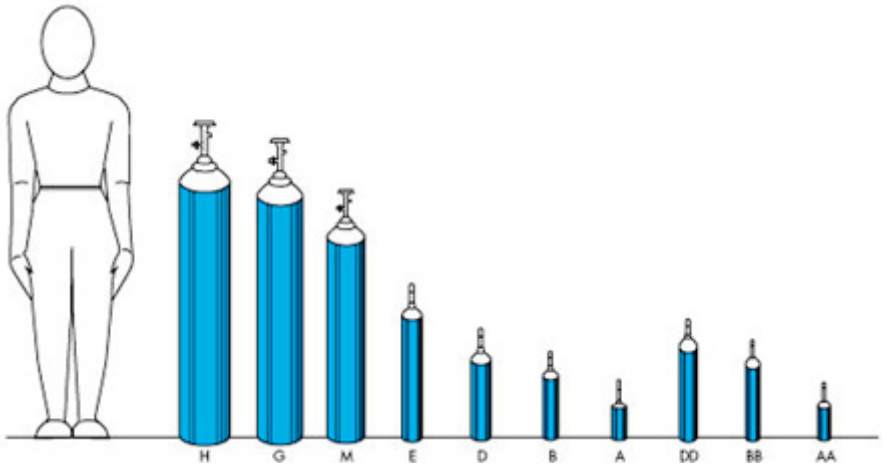


Figure 9: Various **types of high pressure cylinders** used in medical gas therapy. [20]

The portable oxygen is prescribed to patients in LTOT to allow their mobility and let them leave home on a regular basis. To be portable, the system must be lightweight, compact and also capable of providing oxygen for extended periods. [9]

The portable oxygen system consists of a pressurize cylinder/tank, with an on/off valve, a pressure gauge and a flow regulator knob. [21] The Regulator is attached to the top of the tank and is responsible for the oxygen flow control and also for the reducing of the oxygen's pressure to a level that can be safety used by the patient, which is known as the working pressure. [22] It consists of two gauges, one is the pressure gauge that indicates and registers the amount of oxygen that is in the cylinder and the other one is a flowmeter that shows how much oxygen is flowing from the cylinder to the patient and it's marked in lpm. [23] The regulator has attached a flow regulator knob to adjust the liter flow; it determines the rate of oxygen that leaves the tank and goes to the patient. [24]

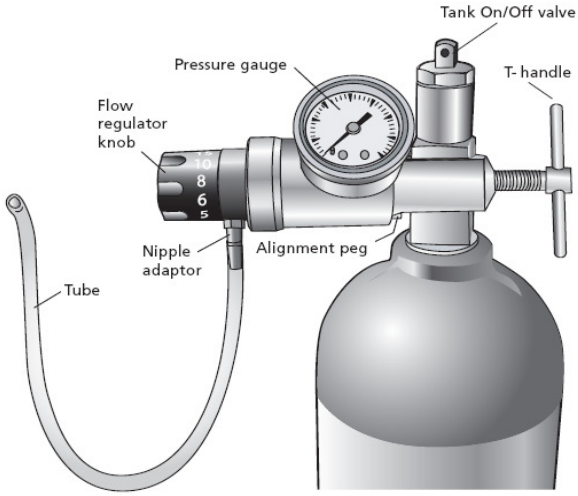


Figure 10: Schematic representation of an **Oxygen Delivery System** constituted by an oxygen cylinder and associated setup: on/ off valve, a pressure gauge and flow regulator knob. [25]

Commonly, the cylinder's valve is opened either with a special wrench or with a handle. As well, the valve has two holes for alignment with the two pins on the oxygen's regulator, which is called the Pin Index System. [26]

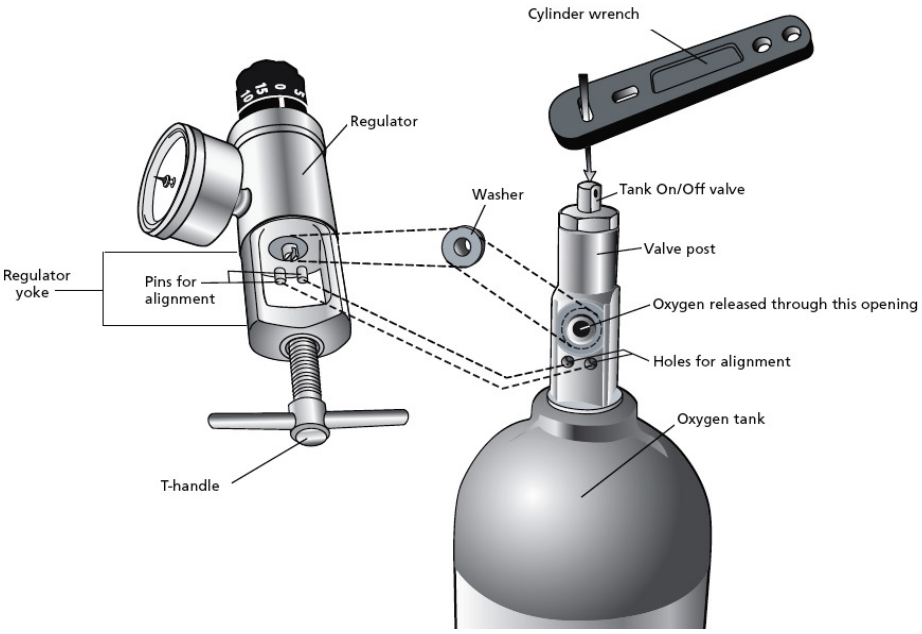


Figure 11: Schematic representation of an **Oxygen Delivery System** showing the pin index system used to attach the regulator to the oxygen tank; the pins for alignment in the regulator match with the holes of the valve. [25]

The Pin Index System's goal is to prevent accidents, ensuring that a medical oxygen tank only receives a regulator for medical oxygen and certifying that only medicinal oxygen is administered to patients in oxygen therapy.

The system consists of a series of protruding pins in the regulator that are in configuration with the associated cylinder outlets, which has a series of pin inlets. These inlets will only allow the regulator designed for that tank to be fitted properly. [24]

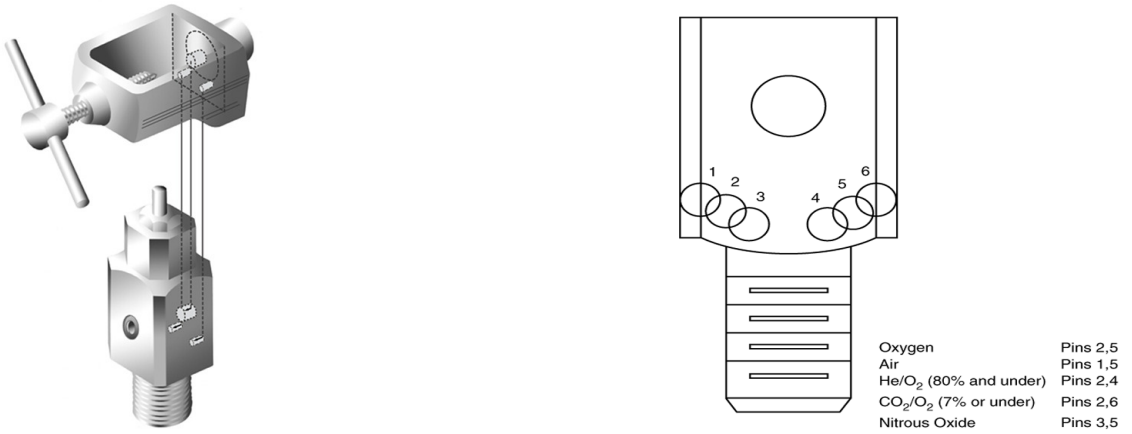


Figure 12: Pin Index System, showing the outlet pins in the regulator and the correspondent inlet pins in the cylinder. The regulator has several pins which are used for specific types of gases. [20]

3.5. Patient Interfaces

To the oxygen's administration can be use several patient interfaces, each one with a different purpose. [16] In the next table there are shown some of these devices, as well as the flow and FiO_2 provided and also the appropriate instance of use.

Device	Flow (lpm)	FiO_2 (%)	Appropriate Use
Nasal Cannula	¼ -8	22-45%	LTOT patients
Transtracheal Catheter	¼-4	22-45%	Patients who don't accept cannula, high flow needs
Reservoir Cannula	¼-4	22-35%	LTOT patients
Simple Mask	6-12	35-60%	Acute short-term therapy requiring moderate FiO_2
Reservoir Mask	6-10	35-60%	Emergencies, acute hypoxemia, moderate FiO_2
Nonrebreather Mask	10-15	80-100%	Emergencies, Respiratory failure

Table 4: Types of **patient interfaces**, the correspondent flow and FiO_2 provided and also the appropriated instance of use. [16]

The patient interface can be classified either in low or high flow device. In Low Flow devices, the oxygen flow provided doesn't match the entire patient's flow needs; therefore, the oxygen given to the patient is mixed with room air. The oxygen's concentration can be either high or low. High Flow devices provide a fixed gas mixture, which meet or even exceed the patient's requirements. The low flow devices are the nasal cannula, the nasal and transtracheal catheters, the simple face mask and the

partial rebreather mask, while the high flow devices are the non- rebreather mask, the Venturi mask and the aerosol systems. [27], [28]

The **Nasal Cannula** is frequently used in home oxygen therapy due to its simplicity and patient convenience. It is composed by two thin plastic tubes, the nasal prongs, which are inserted into the patient's nostrils, providing the entrance of the air. [29] When the delivered flow is higher than 4 lpm it's recommended a humidifiers' use to reduce the patient's discomfort and dehydration of the nasal passages. [27], [28]

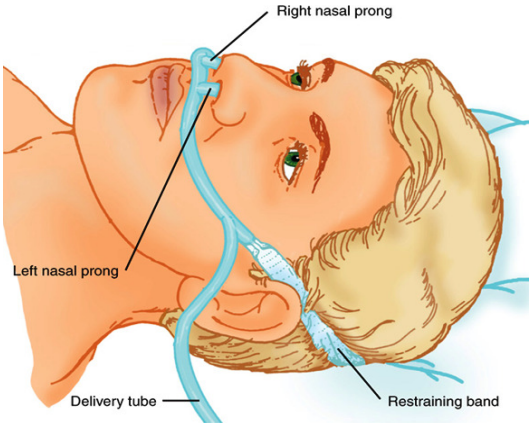


Figure 13: Nasal Cannula, with two nasal prongs insert in the patient's nostrils. [30]

The **Nasal Catheter** is rarely used nowadays; it's uncomfortable and has been replaced by the nasal cannula. [30] It has small holes at the tips which are lubricated with liquid paraffin, to avoid nasal passage trauma, and it's inserted through the nostrils to the oropharynx and is placed behind the uvula. [6] The nasopharynx and the oropharynx act as natural reservoirs, which increase the device's efficiency, nevertheless the nasal catheter can cause gastric distension or even rupture if the catheter is pushed in to far. [31] This device is used to deliver low flows of oxygen, ranging from 0.25-8 lpm with a concentration of 22 to 45%. [30]

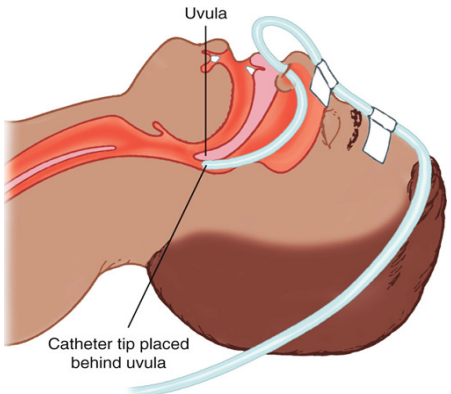


Figure 14: Nasal Catheter with the tip located behind the patient's uvula. [30]

The **Simple Mask** is a face piece that covers the patient's mouth and nose; it has a central port to provide the oxygen's entry through the oxygen inlet connection. The gases are then exhausted through the exhalation ports at the sides of the

mask.[31], [32] The final FiO_2 is dependent of the amount of room air that mixes with the oxygen into the mask, which varies with the patient's respiratory pattern, changing with the peak inspiratory flow rate and also with the respiratory rate. [29], [31]

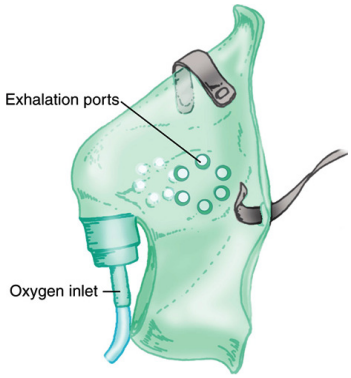


Figure 15: Simple mask, with the lateral exhalation ports and the oxygen inlet to the oxygen's entry. [30]

The **Partial Rebreathing Mask** is similar to the simple face mask, but it has in addition a reservoir bag to collect oxygen during the expiration. This mask's purpose is to conserve oxygen, since some of the exhaled air is rebreathe, reducing the wastes. On the other hand, it can lead to hyperoxygenation, particularly in COPD patients and also dehydration when used with high flows or even eye irritation, with the mask's wrong placement. [32] The FiO_2 achieved is about 0.70 and the flow varies from 6-10 lpm. [31]

The **Non-Rebreathe Mask** has a reservoir bag attached and uses one-way valves on the exhalation port and between the reservoir bag and the mask to deliver the highest possible oxygen's concentration. [31] The attached reservoir bag fills with oxygen between breaths and the valves prevent the inhalation of room or exhaled air through the exhalation ports. [29] The difference between this mask and a partial-rebreather mask is the placement of a one-way valve between the reservoir bag and the mask. [32] This valve prevents the entrance of the exhaled gases in the reservoir bag during the expiration. [31]

Figure 16: Non-Rebreathe Mask with and without the valves placed. [30]

The ***Venturi* Masks** are high flow systems that delivery a fixed FiO_2 to the patient using the *Venturi Effect* as mechanism of actuation. The predetermined concentration of oxygen is delivered into the mask through a jet orifice which is narrow. Because of the high velocity of the oxygen flow and the narrow orifice into the *Venturi* barrel, there's a progressive reducing of the pressure throughout the constricted section. It results in a negative pressure that pulls the room air into the mask through the air entrainment ports. The expired gas is then exhausted through the exhalation holes on the mask to reduce to minimal its rebreathing. The *Venturi* mask can deliver flows ranging from 6-10 lpm with oxygen's concentration varying from 24 to 44%. [31], [32]

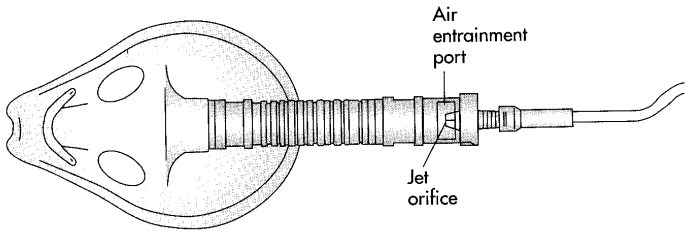


Figure 17: Venturi Mask, with the jet orifice to the oxygen delivery and the air entrainment port to the room air admission. [30]

3.6. Oxygen Conserving Devices

Oxygen conserving devices allow the oxygen's saving, substantially extending the duration of liquid portable or small high-pressure gas systems, making easier the patient's ambling outside home. [33] There are currently available three methods of oxygen's conservation: Transtracheal Oxygen Catheters (TOC), Reservoir Systems and Demand Oxygen Delivery Systems (DODS). [34]

3.6.1. Transtracheal Oxygen Catheter

TOC are plastic tubes, small and flexible insert directly into the trachea to release oxygen. Oxygen flow rates can be reduced in 50% at rest and 30% during exercise, when compared with oxygen delivered by nasal cannulae and for the same SaO_2. [35] TOC has low flow requirements since the anatomical dead space acts like a reservoir, leading to oxygen's conserving. As well, the aesthetics question is also improved and also the taste and smell; in addition, it reduces the soreness that commonly occurs in the face caused by nasal cannulae. [36]

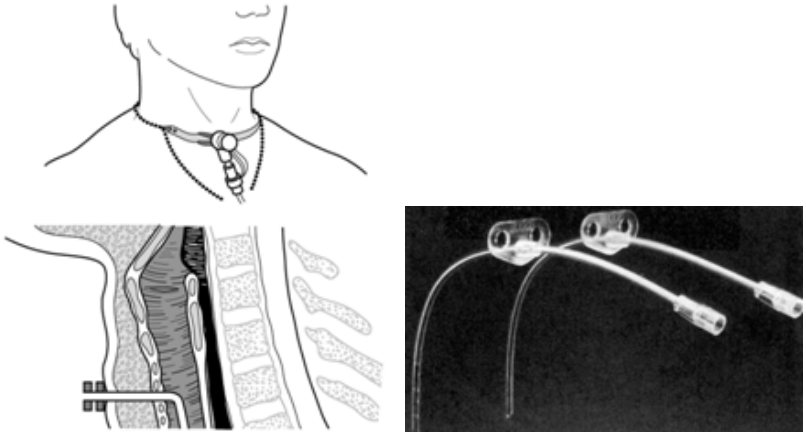


Figure 18: Diagram of the **transtracheal catheter** positioning and a picture of two TOC. [37]

3.6.2. Reservoir Systems

Reservoir systems have an additional reservoir near the airway, in general with a capacity of 20 mL to conserve the oxygen delivered during the exhalation, which reduces the required flow rate and saves a significant amount of oxygen. [36] The oxygen conserved is delivered during the next inhalation as a bolus. [38] The reservoir can be either moustache or pendant type; the moustache type has the reservoir under the patient's nose, while the pendant type is suspended in the front of the patient's chest. [36]

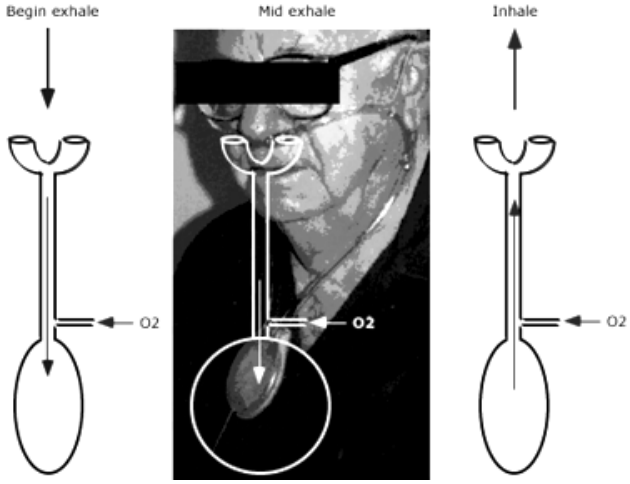


Figure 19: A pendant type reservoir system, the oxygen is stored in the reservoir during the exhalation and rebreathe in the next inhalation. [39]

The reservoir cannula is a moustache type system; it has a small reservoir to collect oxygen during the exhalation to deliver in the inhalation, thus saving oxygen that would be waste to the atmosphere. [34] In *Oxymizer* reservoir nasal cannula (*Chad Therapeutics*, USA), the gas saved during the exhalation is then inhaled at the beginning of the next inhalation, due to the presence of a membrane that creates a

capture chamber. In the exhalation, the membrane is pushed forward to create a capture chamber for oxygen, while in the inhalation the membrane collapses to deliver the collected oxygen. [38]

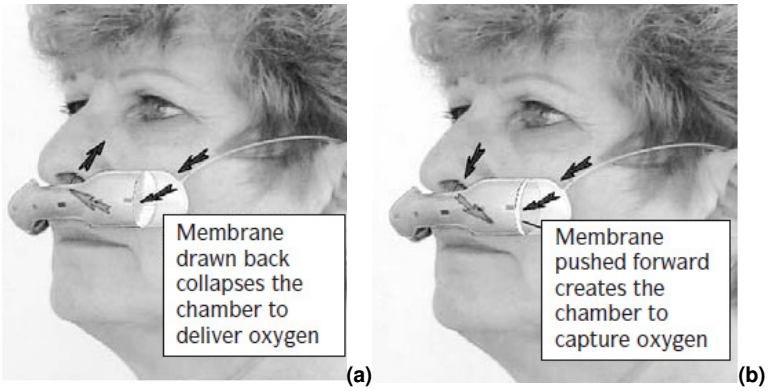


Figure 20: The *Oxymizer* reservoir nasal cannula, during (a) the exhalation and (b) the inhalation. [38]

Reservoir systems are simple, safe and efficient, although the moustache type isn't aesthetics agreeable to the patient that commonly uses it only at home. The pendant type is less perceptible, so it is used outside home more regularly. [36] It has the incontinent of the weight that can cause discomfort in the user's ears loop. Both of the devices work as standard nasal cannulae. [38]

3.6.3. Demand Oxygen Delivery Systems

DODS are conserving systems since they sense the beginning of a breath, which opens the valve, and only deliver oxygen during the inhalation, interrupting the flow during the exhalation, when it would be wasted. [40] These devices are of three types: pulse, demand or hybrid and the difference between each other is in the way how oxygen is delivered. In pulse devices, the oxygen is only provided in the initial part of the inhalation as a bolus of gas, but in higher flow rates than in continuous flow. The demand devices provide a bolus of oxygen at the beginning of the inhalation and maintain a continuous and smaller flow during the entire inhalation, so the flow rate needed is less than for the pulse devices. [36], [40] The hybrid device joins the operational principles from both of pulse and demand devices, providing an early bolus in the onset of the inhalation, followed by a continuous flow. [41]

DODS can be classified as electronic or pneumatic/mechanical according to the technique that they use to open the valve. Pneumatic conservers use the gas pressure to actuate the valve and they don't require electric power. Electronic conservers are battery powered and use the time predetermined by the manufacturer to open the valve. [36]

Figure 21: Intermittent-flow oxygen-conserving devices. Front row (left to right): EX-3000 (DeVilbiss, Somerset, Pennsylvania), EX-2000 (DeVilbiss), Impulse Select (AirSep, Buffalo, New York), CR-50 (Mallinckrodt, St Louis, Missouri) OxiClip (Mallinckrodt). Back row (left to right): O2N Demand (Victor, Denton, Texas), O2 Advantage (Western Medica, Westlake, Ohio), Venture (Invacare, Elyria, Ohio), DOC-2000 (Transtracheal Systems, Englewood, Colorado), Oxymatic 301 (Chad Therapeutics, Chatsworth, California). [36]

Although oxygen conserving devices enable significant oxygen's savings, not all the patients accept these apparatus; when they pass from continuous flow oxygen therapy to an intermittent flow oxygen therapy, some patients do not become accustomed and fell like they're not receiving enough oxygen, which causes them discomfort.

3.7. Oxygen Dosing

Commonly, when it's used the oxygen delivery with continuous flow, the dosing prescribed is 2 lpm through nasal cannula. This fixed amount results from two studies made in the 60's, one in Winnipeg, Canada, and another one in Denver, USA. In both of them, there was analysed a small amount of COPD hospitalized patients whose oxygen flow rate was increased gradually, under state stable conditions. Both of them demonstrated that the COPD patients can be managed with a fixed flow of 2 lpm. So, the Nocturnal Oxygen Therapy Trial (NOTT) established for patients with COPD and hypoxemia the oxygen delivering of 1 to 2 lpm by nasal cannula and during sleep and exercise the flow is increased 1 litre, from 2 lpm to 3 lpm, due to the common the desaturation that occurs in the sleep hours, specially in the REM phase of sleep. [42]

3.8. Pulse Oximetry

Pulse oximetry applies the optical properties of blood to continuously and non-invasively monitor the level of oxygen in the blood stream and the heart rate. The pulse oximeter has a pair of LED's that operate at two different wavelengths: one is a red LED with a wavelength of 660 nm and the other is a near infrared LED with a wavelength of 910 nm. Both of the LED's are placed in the opposite way of a photodiode, an optical sensor that detects the emitted light from each LED. The sensor is placed either in the finger or in the ear. [43]

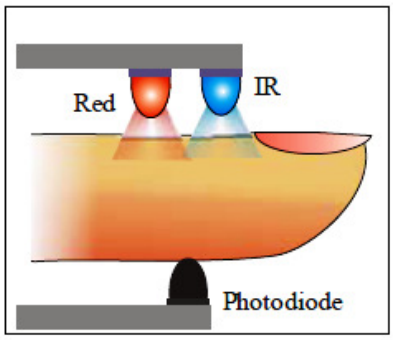


Figure 22: Representation of a typical **pulse oximeter**, the two LED's, red and near infrared, emit light that passes through the patient's finger and is then collected by the photodiode. [43]

This technique is based in the blood's colour change due to the variable quantity of light absorbed by the haemoglobin, which depends on its saturation with oxygen. When the haemoglobin is oxygenated, constituting the oxyhaemoglobin, it doesn't absorb large quantities of red light, but when haemoglobin oxygen's saturation falls, this quantity increases, turning the blood darker. On the other hand, the near infrared light is more absorbed by oxyhaemoglobin than by haemoglobin. [44] Accordingly, the absorption coefficients are different for the two types of haemoglobin and also vary with the light wavelength, as represented in the next figure:

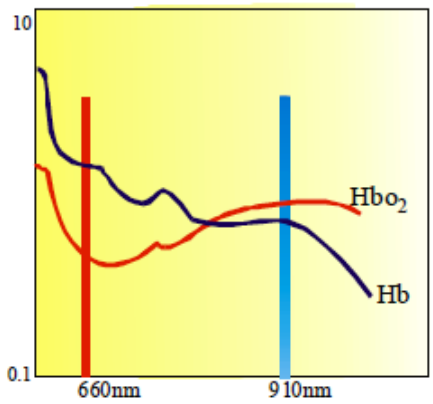


Figure 23: Absorption coefficient for the two types of haemoglobin for each LED's wavelength; the red light (660 nm) is more absorbed by the haemoglobin and the near infrared light (910 nm) is more absorbed by the oxyhaemoglobin. [43]

The blood oxygen saturation, SpO_2, is calculated based on Beer-Lambert law, which uses the relation between the concentration of a solute and the intensity of light transmitted through a solution at a specific wavelength. According to this law, the concentration of a solute in a solvent is determined by the amount of light that is absorbed by the solute at a specific wavelength; as oxyhaemoglobin and deoxyhaemoglobin preferentially absorb one of the two wavelengths of light used, their concentrations can be simply determinated. [44] Thus, SpO_2 is defined as the ratio of oxyhaemoglobin to the total concentration of haemoglobin, as represented in the equation below:

$$SpO_2 = \frac{[HbO_2]}{[Hb + HbO_2]}$$

Equation 1: The blood oxygen saturation, SpO_2, is given by the ration between the concentrations of oxyhaemoglobin and the total of haemoglobin.

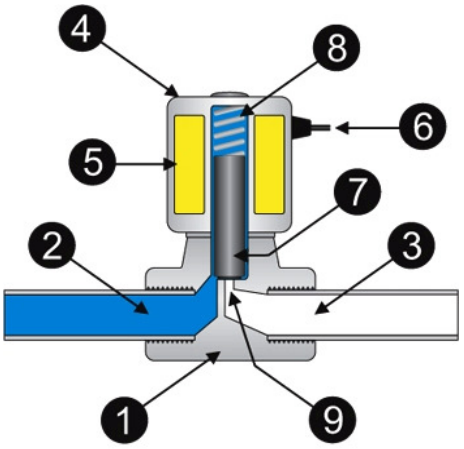
The pulsatile change in light transmission is caused by variations in arterial blood volume between the source and the light detector and represents the blood pulsing in the arteries. The blood pressure varies in the heart cycle and the amount of blood in the capillaries depends directly on it; so the heart rate can also be measured through this method. Hence, this technique is a combination of two technologies of spectrophotometry, which are responsible for the blood oxygen saturation measurement and optical plethysmography, which measures the pulsatile changes in arterial blood volume at the sensor placement. [44]

3.9. Electronic Proportional Valve

An electronic proportional valve is a mechanical device electromagnetically-operated that is commonly used in closed-loop systems to control the flow or pressure rate. It's a type of solenoid valve; however in solenoid valves the electrical current is just switched on/off while in proportional valves the current is regulated to accurately control the output flow/pressure. These equipments are termed proportional since the output flow/pressure isn't exactly linear in relation to the input current. [45], [46]

The proportional valve is composed by two main parts: the solenoid and the valve. The solenoid is a coil of wire that becomes magnetized when electricity is run through it and it's responsible for electrical energy's conversion into mechanical energy, which opens and closes the valve mechanically. The valve also has a spring, which is an elastic object that stores mechanical energy and is commonly made of hardened steel. It is responsible for maintaining the valve either opened or closed while

it isn't activated. The valve has at least two ports, one inlet and one outlet port, to the inlet and outlet flow, but it can also have three or even more ports.



1. Valve body
2. Inlet port
3. Outlet port
4. Coil/Solenoid
5. Coil windings
6. Lead wires
7. Plunger
8. Spring
9. Orifice

Figure 24: Solenoid Valve, normally closed type, and its components. [47]

The valve's principal of operation is the balanced between two proportionally and opposite forces acting on the plunger: the mechanical force provided by the spring and the magnetic force, created by the electrical current passing through the solenoid. The electrical current flowing through the coil creates a magnetic field which is proportional to the current applied. Therefore, the force produced by the solenoid results from strength of the magnetic field, which forces the plunger down. The mechanic force applied by the spring forces the plunger up. As the input current varies, the output opening varies as well, resulting in a wide variety of output diameters. So, by varying the current in the solenoid, the amount of coil movement can be varied and hence the amount of flow through the valve can be controlled. [45], [47]

The valve can be either normally closed or normally open. In normally closed valves, they open when the coil is magnetized and re-close when the current is removed from the coil. In normally open valves, when the current flows through the coil, it becomes energized, closing the valve. When there isn't applied any electrical power the valve is therefore opened. [47]

The valve can be controlled by varying the voltage as well as the electrical current to the solenoid. However, varying the voltage it isn't as accurate as varying the current, since at a certain voltage the correspondent current increases the resistance of the coil by temperature rise. The current determines the magnetic force that opens the valve, so at the end, the flow through the valve may be reduced because the valve will tend to slowly close after a while when the control is done by voltage varying. The deviation is about 4% in voltage control versus current control. [48]

3.10. Proportional, Integral, Derivative Controller

The Proportional, Integral, Derivative (PID) control is extensively used in systems that involve automatic management. Its structures may be modified, but its bases remain the same. PID algorithm principle's is the feedback control; it has the ability to control the output of a process variable, approaching and maintaining it from a desired value, the setpoint. The PID controller measures the process variable and compares it to the setpoint. According to the control error, the PID controller calculates the correction to be applied in the control device, which changes the process input. Those corrections will allow the process to become equal or within the range of the setpoint. Therefore, the PID algorithm involves three different terms, the Proportional, the Integral and the Derivative term, each one with a specific purpose. [49] The algorithm's is described by follow equation:

$$u(t) = K_P(e(t) + \frac{1}{T_i}\int_0^t e(\tau)d\tau + T_d\frac{de(t)}{dt})$$

Equation 2: PID algorithm's equation, where u(t) is the control signal, e(t) is the control error, K_p is the proportional gain, T_i is the integral time and T_D is the derivative time.

$$u(t) = K_P e(t) + K_i\int_0^t e(\tau)d\tau + K_D\frac{de(t)}{dt}$$

Equation 3: PID algorithm's equation, where u(t) is the control signal, e(t) is the control error, K_p is the proportional gain, K_i is the integral gain and K_D is the derivative gain.

Where:

$$K_i = \frac{K_P}{T_i} \qquad \textbf{Equation 4}$$

$$K_D = K_P \times T_D \qquad \textbf{Equation 5}$$

u(t) is the control signal and e(t) is the control error, given by the difference between the setpoint and the measured value. If the measured variable is y and the setpoint is y_{sp}, the error is:

$$e(t) = y_{sp} - y$$

Equation 6: Control error, e(t), where y is the measured variable and y_{sp} is the setpoint.

The control signal is therefore the sum of three terms, the P-term that is proportional to the error, the I-term that is proportional to the integral of the error and the D-term that is proportional to the derivative of the error. K, T_i and T_d are the controller parameters, which are the proportional gain, the integral time and the derivative time, correspondingly. Those three terms can be view as the past, present and future contributions to the controller. [50]

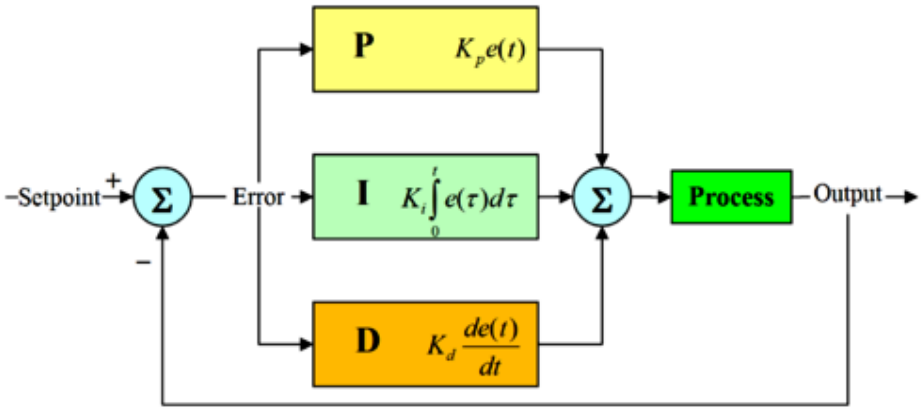


Figure 25: Schematic representation of a **PID controller**. [51]

The PID controller can be used on a microcontroller, which allows the automatic tuning, the gain scheduling and also a continuous adaption. [50]

3.10.1. Proportional Term

The proportional term, also called gain, directly reacts to a non-zero control error to make smaller its contribution to the measured process variable. This term is given by the equation:

$$P = K_P e(t)$$

Equation 7: Proportional term from the PID controller, where P is the proportional term of the output, K_P is the proportional gain and e(t) is the control error.

The effect of this term is always prevailing when the control error is large and is reduced when the error approaches to zero. [49]

A high K_P leads to a large change in the output for a certain change in the error, but for K_P values too high, the system's response can become unstable since the tendency to oscillate is increased. Nevertheless, small gains result in small changes in the output response due to a large input error, which results in a less sensitive controller to system disturbances. [49], [51]

3.10.2. Integral Term

The integral term, also called reset, progressively removes the control error, correcting the offset and the strength of this term is increased with decreasing integral times (T_i). [49] It is proportional to both of the error and its duration. [51] This term is given by the equation:

$$I = K_i \int_0^t e(\tau)d\tau$$

Equation 8: Integral term from the PID controller, where I is the integral term of output, K_i is the integral parameter and e(τ) is the control error.

The integral control increases the movement of the process towards the setpoint, but, since the integral term responds to the past's accumulated errors, the present value can overshoot the setpoint, crossing it in other direction. [51]

3.10.3. Derivative Term

The derivative term reacts to the change rate in the process variable: if it changes considerably, the derivative control reacts robustly. This term is given by the equation:

$$D = K_d \frac{de(t)}{dt}$$

Equation 9: Derivative term from the PID controller, where D is the derivative term of output, K_d is the derivative coefficient and e(t) is the control error.

The derivative term controls the rate of change in the controller output, which is more obvious close to the setpoint. This term reduces the overshoot produced by the integral term and improves the system's stability. However, the signal's differentiation amplifies its noise, so this term amplifies the noise of the process variable and it becomes unstable if the noise/derivative gain is too large. So, as differentiation is always sensitive to the noise, it's therefore necessary to apply a low-pass filter, which attenuates the high frequencies, that is, the measurement noise from the process variable. [49], [51]

3.10.4. Tuning

The PID control success's depends on the right choice of the control constants, which is also one of the most difficult steps in its implementation. There are several methods for the parameter's calculation, which are often called tuning methods, such as the manual tuning, the Ziegler-Nichols and the Cohen-Coon methods.

The Ziegler-Nichols methods are very well-known. They are based on the characterization of the process dynamics using a few and simple parameters and equations; they offer good tuning and are simple to apply. The step response method by Ziegler and Nichols is based on the process information, which has the form of an open-loop step response; that response is obtained as in the bump test. [50] The system lag and a general slope illustrated are used to set the PID gains. The generic system open loop step response of a first order system is shown next, as well as the curve parameters [52]:

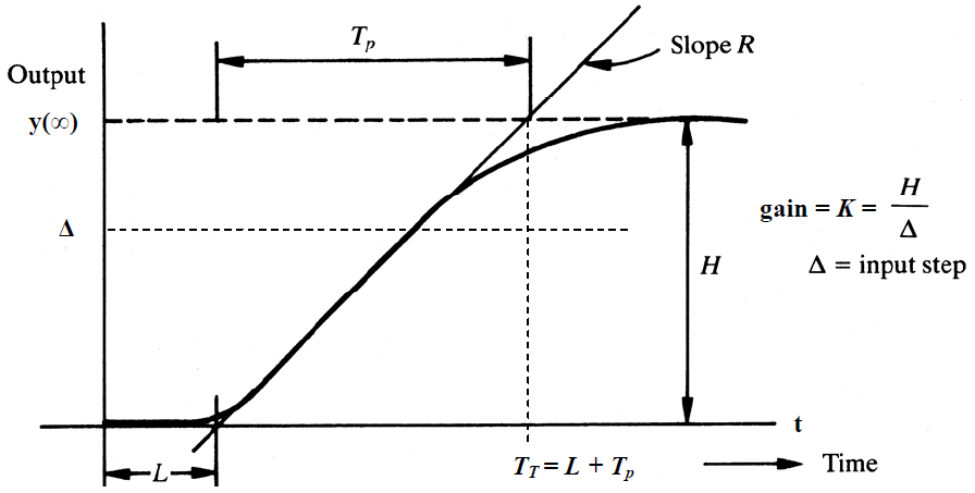


Figure 26: Open loop step response of a first order system. [52]

Based on the open loop response, the PID analogical gains are calculated using the follow equations, where L is the dead time and R is the slope of the curve [52]:

$$K_P = 1.2 * R * L \qquad K_I = \frac{K_P}{2L} \qquad K_D = 0.5 * L * K_P$$

Equation 10: Ziegler-Nichols step response method to obtain the PID analogical gains. [53]

After that, Takahaski suggested the following method to the tuning of a digital PID. The sampling time, T_s, and the controller gains are calculated with the follow equations [53]:

$$K_p = \frac{1.2}{R(L+T_S)} \qquad K_i = \frac{0.6T_S}{R\left(L+\frac{T_S}{2}\right)^2} \qquad K_D = \frac{0.5}{RT_S} \text{ to } \frac{0.6}{RT_S} \qquad T_s = 0.3L$$

Equation 11: PID digital gains. [53]

3.11. Microcontrollers

A microcontroller consists on a single integrated circuit with a central processing unit (CPU) combined with several support functions such as a crystal oscillator, timers, input/output ports, watchdog, and others. Microcontrollers are designed to work alone; they don't need external components, since the peripherals are already built into them, which save time and space. [54]

3.11.1. Memory

The memory's function is to store data and it contains all the memory locations. The input data has a certain address of memory location; so, when this address is selected it accesses the content in that location. Accessing contents in memory also takes a certain time to locate it. Besides reading from a memory location, memory also provide for writing onto it, through an additional line called the control line (R/W (read/write)). This line is used in the following way: if R/W=1, reading is done, and if R/W=0, writing is done on the memory location. [54]

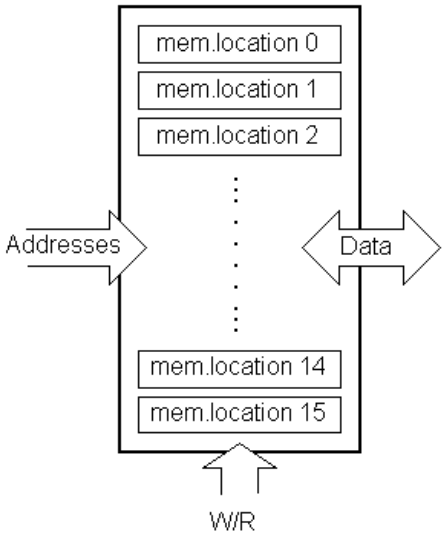


Figure 27: Memory unit, for a specific input there is a correspondent output, the control line R/W determines if we're reading or writing to the memory. [54]

3.11.2. Central Processing Unit

The CPU is constituted by memory locations, called registers, which allow the microcontroller to implement various types of mathematical operations, as multiplying, dividing, adding and subtracting, and other types of operations with stored data from the memory unit. [54]

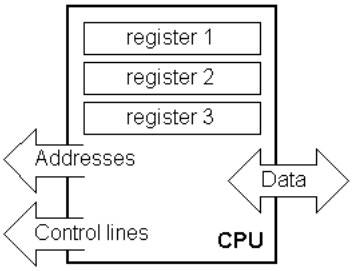


Figure 28: CPU with three registers. [54]

3.11.3. Bus

The bus is a line throughout data goes from one block to another. Consequently, connecting the memory and the CPU using busses gives the microcontroller functionally. The bus is constituted by a group of 8, 16, or more wires. Microcontrollers have two types of buses: address and data. The address bus is composed by as many lines as the amount of memory that is addressed and it's used to transmit addresses from the CPU to the memory unit. The data bus is as wide as data or the connection line and it's used to connect the blocks inside the microcontroller. [54]

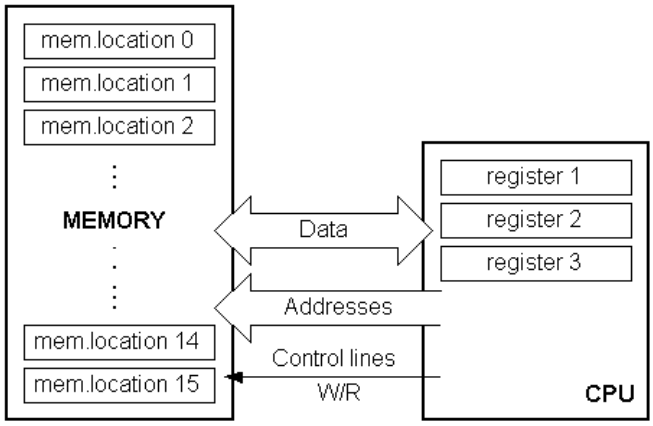


Figure 29: Address and Data Bus, connecting the memory unit and the CPU. [54]

3.11.4. Input/Output Unit

Input/Output unit is used to communicate with the outside devices and also to gate information. This block is constituted by memory locations where one end is connected to the data bus, and the other is connected to the output lines, which are the pins of the microcontroller. These locations are called ports, which can be input, output or even bidirectional ports. Through the ports, it can be read and be written information. [54]

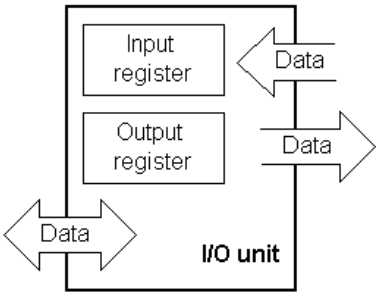


Figure 30: Input/Output Unit. [54]

3.11.5. Serial Communication

Serial communication is a process of communication where data is send as one bit at one time sequentially over a communication channel. It is sent a start signal prior to each byte and a stop signal after it. The start signal is used to prepare the information's reception, while the stop signal brings the receiving to rest in preparation for the next byte's reception. Serial communication differs from parallel communication where several bits are sent together through several parallel channels. [55]

The serial communication block enables the full-duplex communication, where there are separated lines to receive and send information, which can occur at the same time. After receiving the information, it is read from the receiving location and stored in the memory. To send information, data goes from the memory through the bus to the sending location and then to the receiving unit.

The communication obeys to a communication protocol, which is a group of rules for exchanging data. [54]

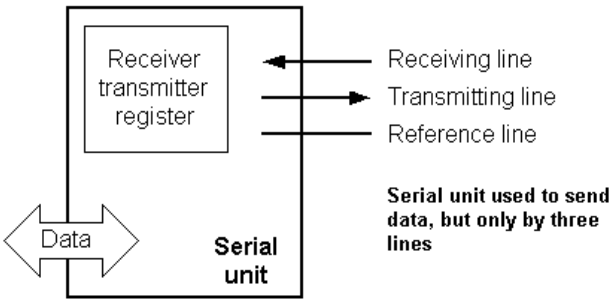


Figure 31: Serial communication block. [54]

3.11.6. Timer Unit

The timer unit gives helpful information about time and duration. It's composed by a free-run counter, which is a register. Therefore, the register value is periodically incremented by one in constant intervals. [54]

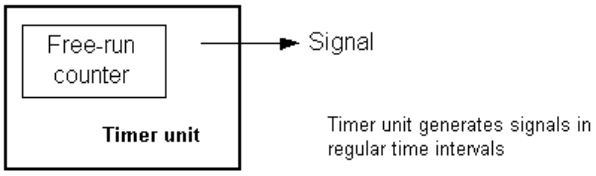


Figure 32: Timer Unit. [54]

3.11.7. Watchdog

The watchdog block is a free-run counter where the program writes “0” every time it executes correctly. If occurs any problem in the program execution, the watchdog won’t write “0” and the counter will reset the microcontroller when it gets to its maximum value. Then, the program is executed again and correctly. [54]

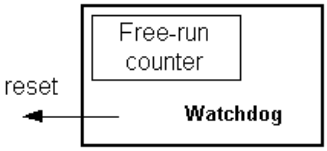


Figure 33: Watchdog. [54]

3.11.8. Analogue to Digital Converter

The outside signals are usually different from the ones in microcontrollers; they often are analogue, and the microcontroller can only recognize digital signals, which are either “0” or “1”. So, external signals must be converted from their analogue value to a digital number that can be processed by the CPU. This task is executed by the analogue to digital converter block. [54]

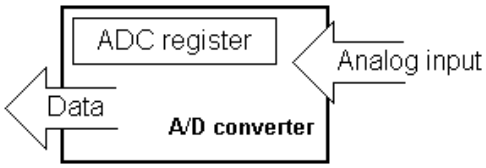


Figure 34: Analogue to Digital Converter. [54]

3.12. Microcontroller PIC24HJ128GP306

The microcontroller used in the system is the 16-bit microcontroller PIC24HJ128GP306 from Microchip®. In this area, the student focused the Theoretical background in the specific modules involved in the valve's integration and actuation.

Figure 35: PIC24HJ128GP306, from Microchip®. [56]

3.12.1. Timers

The PIC24 has nine 16-bits timer modules (Timer1 to Timer9), which have the same functional circuitry but they also have some differences, so they're classified in three types: Type A (Timer1), Type B (Timer2, Timer4, Timer6, Timer8) and Type C (Timer3, Timer5, Timer7, Timer9). Type B and Time C timers can be concatenated to compose 32-bit timers. The Timer module supports different modes of operation, which are the timer mode, the gated timer mode, the synchronous counter mode and asynchronous counter mode, which is only available in Type A timer. In timer and gated timer modes, the input clock is used the internal instruction cycle clock, while in synchronous and asynchronous counter modes, the input clock is derived from the external clock input at the TxCK pin. [57]

3.12.2. UART

The PIC24 also contains two Universal Asynchronous Receiver Transmitter (UART) modules, UART1 and UART2. UART is a module of serial input/ output that enables serial communication with peripheral devices, such as personal computers, and uses communication protocols like RS-232 and RS-485. This module is full-duplex, with 8 or 9 bit data communication, receiving and transmitting data through UxRx and UxTx pins of the PIC, correspondingly. The UART uses the Non-return-to-zero protocol to communicate, which is 1 start bit, 8 or 9 data bits and 1or 2 stop bits. The UART1 uses RS-485 as input/output, while UART2 uses RS-232. [57]

3.12.3. Pulse Width Modulation

Pulse Width Modulation (PWM) of a signal or a power source is a technique for controlling analogue circuits with a processor's digital output and it's used in a large variety of applications, such as communications, conversion and power control due to its low power, noise-free and low cost. [58], [59]

The PWM's control uses a square wave whose duty cycle is modulated. The duty cycle is the amount of time in the period that the pulse is high and usually is specified as a percentage of the pulse period, as indicated in equation 2.

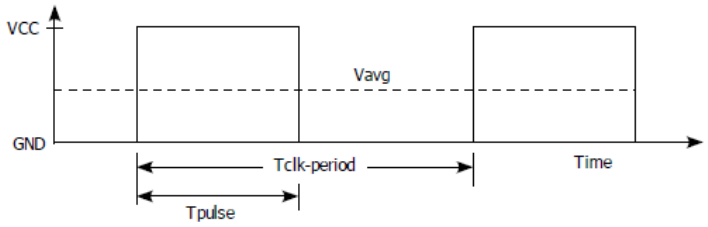


Figure 36: A square wave with 50% of duty cycle. [59]

$$Duty\ \ Cycle\ = \frac{T_{pulse}}{T_{clk-period}} \times 100$$

Equation 12: Wave's duty cycle.

$$Average\ Voltage = Duty\ \ Cycle \times V_{CC}$$

Equation 13: Wave's average voltage.

By varying the square wave's duty cycle its width is modulated resulting in the variation of the average value of the waveform and generating a variable output voltage, by means of a repeating series of on and off pulses. [59], [60]

In the next figure, there are represented three different PWM signals. The first wave has a PWM output at a 20% duty cycle, so the signal is on for 20% of the period and off the other 80% of the time. The second and third waves have PWM outputs at 50% and 90% duty cycles, correspondingly.

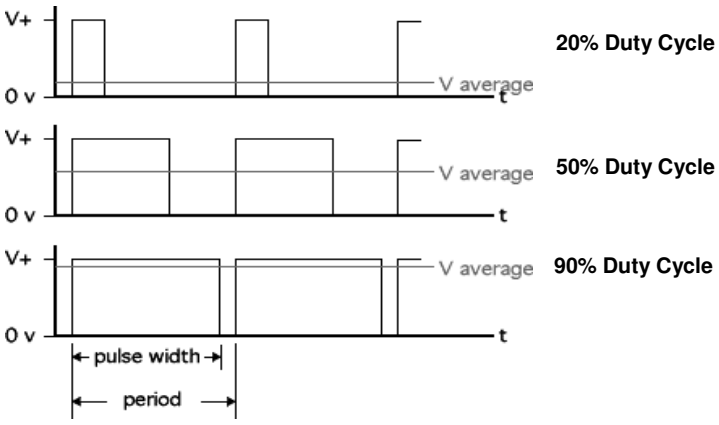


Figure 37: PWM signals with different duty cycles, the signal is in high level for 20%, 50% and 90% of the period respectively. [61]

The PWM signal is digital because, at a certain instant of time, the signal is either completely on or completely off. [58], [59]

Microcontrollers have PWM outputs using a counter, the timers, which is periodically incremented by a clock signal internal or external to the digital circuit. When the counter period is equal to the reference value, the PWM output changes state; the counter reset when it gets to the PWM's period. The duty cycle is therefore varied in discrete steps and it's a function of the timer resolution, which is the highest number of pulses that can be packed in the PWM period. [60]

3.12.4. Output Compare

The Output Compare module has several associated operational modes. Generally, this module compares the value of the timer with the value of one or two compare registers depending on the operating mode selected. Then, the state of the output pin changes when the timer value matches the compare register value. The output compare module generates either a single output pulse or a sequence of output pulses, by changing the state of the output pin on the compare match events.

To control the voltage applied in the valve's solenoid it was used the PWM mode without fault protection of this module. This mode is similar to PWM mode with fault protection, but this one can be activated externally. This module uses a timer which has to be Timer2 or Timer3; this resource is selected by configuring the Output Compare Timer Select (OCTSEL) bit in the Output Compare Control register.

The PWM mode is used to generate variable duty cycle output. The selected timer starts counting from zero and increments on every clock until it reaches the value in the Period Register (PRy). The Compare Register (OCxR) value is continuously compared with the timer value and when the match occurs, the OCx pin is driven low. When the period value is reached, the timer resets to zero and starts incrementing once again. The timer can be clocked using an internal clock source ($F_{OSC}/2$) or a synchronized external clock source applied at the TxCK pin. The PWM duty cycle is specified through Secondary Output Compare (OCxRS) register, and the Output Compare (OCxR) register is a read-only compare register. [57]

$$PWM\ \ Period = (PRy + 1) \times TCy \times (TMRy\ \ \mathrm{Pr}escale\ Value)$$

Equation 14: PWM period calculation.

$$FCy = \frac{F_{osc}}{2}$$ **Equation 15**

$$TCy = \frac{1}{FCy}$$ **Equation 16**

4. State of the Art

There isn't available in the market any device that controls in real time and automatically the oxygen flow to the patient, based on his real needs and that can prevent waste of oxygen. The equipments available only enable the manual control. Nevertheless, there is under development a system to achieve that goal, which is near to the one we proposed to build up in many points. There are also studies which confirm the efficiency of the saturation driven oxygen therapy, where the patient's SaO_2 is used to control the oxygen's delivery, establishing a dynamic therapy. There are also some studies to implement a PID controller in mechanical ventilators and also in neo-natal units to maintain the patient's oxygen saturation as close as possible from a pre-defined value defined by the clinicians and more accurately control its variation. Those studies are mentioned later in chapter 6.

4.1. Saturation Driven Oxygen Therapy

The Department of Bioengineering, South Kensington Campus of London has simulated a method to automatically regulate the oxygen flow rate in response to the measured oxygen demand by pulse oximetry, which is called the Saturation Driven Oxygen Therapy (SDOT). This type of treatment uses the oximetry measurements to accurately control the flow as a feedback system, on that way, the therapy matches the patient needs to oxygen supply more exactly than in the conventional fixed flow therapy. The system simulated consisted in a controller that receives the pulse oximeter measurements of oxygen saturation and computes the error (E(t)) between the target setpoint and the oximetry value measured. Then, the controller determines the optimum oxygen flow rate and outputs a signal to a flow regulator, managing the delivery of oxygen to the patient. The feedback from the pulse oximeter enables the controller to handle the oxygen flow, since there is feedback of SpO_2 to the closed-loop controller to safer control the output. Furthermore, the microcontroller is programmed with the PID algorithm to compute the correct dose of oxygen. [2]

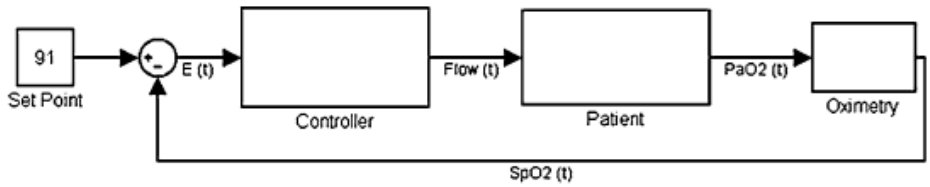


Figure 38: Block diagram of the system tested. [2]

To the patient safety, the oxygen flow had limits, a maximum (5 lpm) and a minimum (0 lpm), which were provided by a non-linear saturation regulator. The group had also developed a mathematical model to modulate the patient blood oxygen saturation response due to the flow's variations, through the study of the PaO_2 variations. Accordingly, the PaO_2 is modelled by the follow equation:

$$PaO_2(t) = P_0 + P_d(t) + P_f(t) \ \ (mmHg)$$

Equation 17: Patient arterial oxygen concentration model. [2]

The equation has three terms; P_o, a baseline level, P_d, an arbitrary disturbance and P_f, the flow rate dependence. Where $P_o \approx 60$ mmHg in the absence of disturbances or any oxygen flow and the arbitrary disturbance is given by:

$$P_d(t) = -3 + 3\cos(2\pi * f * t)$$

Equation 18: Arbitrary disturbances introduced by the variable conditions possible in COPD patients during sleep and exercise. The frequencies used in the simulation varied between 10^{-3} and 10^{-1} Hz. [2]

The patient's oxygen flow dependence, P_f is described by a second order linear differential equation that relates the oxygen flow rate (U) to the arterial oxygen concentration:

$$P_f(t) = \alpha\left[M\frac{d^2U(t-T_1)}{dt^2} + B\frac{dU(t-T_1)}{dt} + KU(t-T_1) \right]$$

Equation 19: Patient's oxygen flow dependence. [2]

α is a sensitivity factor that is for each additional lpm of flow provided, the FiO_2 increases 3% and it was assumed that also the PaO_2 increases 3% with each additional l/min of oxygen.

The pulse oximeter measures the arterial oxygen concentration as a percent of saturation. So, to obtain the arterial oxygen concentration in mmHg it's used the oxyhaemoglobin dissociation curve which equation is given next:

$$PaO_2 = \frac{-\ln\left[1 - (SpO_2)^{0.5}\right]}{0.046}$$

Equation 20: Oxyhaemoglobin dissociation curve. [2]

The system has been simulated on *Simulink*™ (Matworks, Inc, Massachusetts USA) and the results were compared against a simulation of untreated and continuous LTOT patients. The "No Oxygen Therapy" group was composed by three COPD patients where was collected saturation data in the sleeping period using an oximeter during 8 h. Then, this information was incorporated in the simulation, with the previous conversion of the SpO_2 readings to PaO_2 values (using the oxyhaemoglobin

dissociation curve). The SDOT group represents the results of the automatic control of the flow with the controller and the continuous LTOT was also simulated. The results obtained and the comparison of the therapies are summarized in the next table:

Group	Variables	Patient 1	Patient 2	Patient 3
No Oxygen Therapy	Mean overnight SaO_2 (%)	88.8	91.8	95.1
	Standard deviation SaO_2 (%)	4.6	2.3	3.3
	Percent Time vw' SaO_2<90 (%)	55.1	27.3	12.1
	Recording Duration (min)	483	483	483
SDOT	Mean overnight SaO_2 (%)	91	92.5	95.4
	Standard deviation SaO_2 (%)	2.6	1.6	2.7
	Percent Time vw' SaO_2<90 (%)	22.9	4.3	1.8
	Mean flow-rate (L/min)	2.3	0.9	0.4
O_2 equivalent constant flow LTOT	Mean overnight SaO_2 (%)	90.7	92.3	95.3
	Standard deviation SaO_2 (%)	3.7	2.1	3.2
	Percent Time vw' SaO_2<90 (%)	36	15.9	8.5

Table 5: Comparison of the therapies.

The values of SpO_2 obtained were compared with the values from the patients without supplemental oxygen and the SDOT demonstrated that it attenuated the fluctuations, by varying adequately the flow rate. As can be seen in the table above, the SDOT maintained the SpO_2 above the threshold, improving the efficacy of the treatment, when compared to standard LTOT. This simulation demonstrated the potential efficiency of the closed-loop flow control, with an improvement in the oxygen saturation. [2]

4.2. SmartBlender™, Columbia Life Systems

Columbia Life Systems provides devices to the adaptive adjustment of the oxygen delivery to people recovering from respiratory distress. Their device, SmartBlender™, was developed with the contribution of neonatologists and respiratory therapists to modernize the current standard of care worldwide. This device is an oxygen blender, which mixes oxygen with air, controlling the oxygen flow rate and concentration and it automatically controls the amount of oxygen delivered to patients. The SmartBlender™ is composed by a pulse oximeter connected to a minicomputer that is programmed to adaptively adjust the oxygen blender that supplies the oxygen mixture to the face mask or nasal cannula. Consequently, it will adjust more accurately and timely the supplemental oxygen to the patients. The computer also includes an extensive variety of safety features to ensure patient safety. [62]

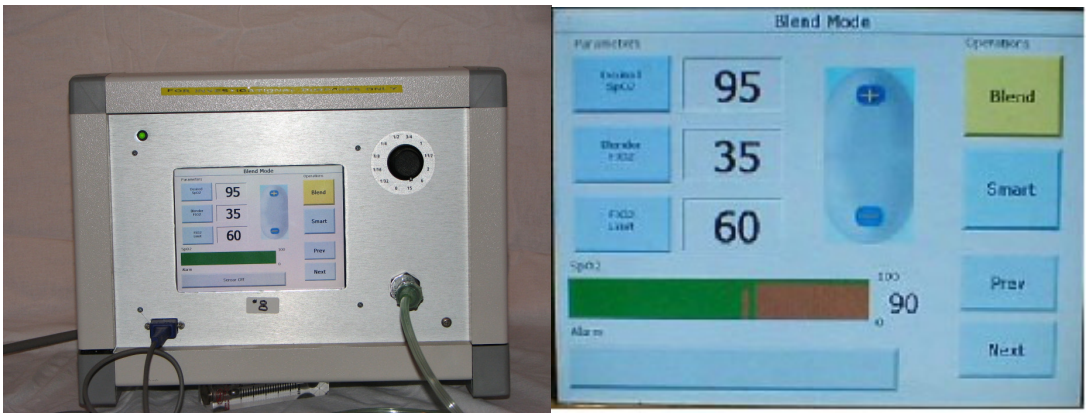


Figure 39: SmartBlender™ and its setup panel. [62]

4.3. Self-Regulating Oxygen Delivery System

The Biomedical Engineering Department from the University of Michigan, U.S.A., in cooperation with the Department of Computer Science from Purdue University, U.S.A., has been working in a project of a Self-Regulating Oxygen Delivery System. The goal of this project was to design a self-regulating device to control the oxygen flow to the patient based on SpO_2 readings and automatically adjust the valve opening based using the pulse oximeter signals. The device also has some features like portability, safety, user-friendly and cost-effectively, thus it could be used either at the hospital or at home. [63]

The system consists in a Freescale HCS12 Microcontroller, an Electronic Proportional Valve (Parker HF Pro Valve), a BCI Pulse Oximeter Sensor and Board, connected in the way represented in the next figure:

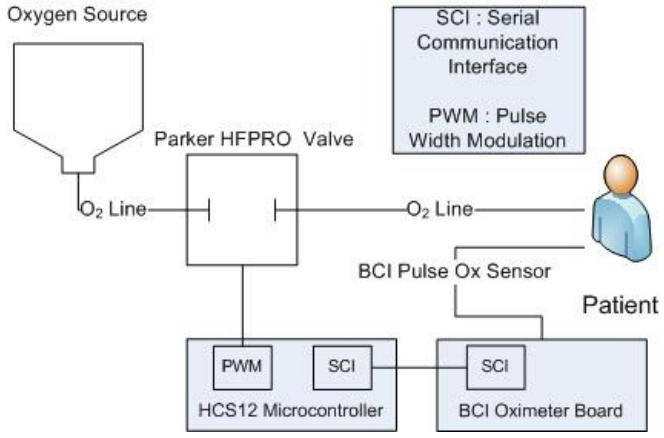


Figure 40: Project Design, representing the connections between the HCS12 Microcontroller, the Electronic Proportional Valve (Parker HF Pro), the BCI Pulse Oximeter Sensor and Board with the patient and the oxygen source. [63]

The BCI pulse oximeter sensor collects and sends data to BCI oximeter board. Then, the BCI oximeter board calculates the SpO_2 and sends a signal to the microcontroller via serial communication interface. The SCI uses an EIA-232 transceiver to enable the communication between the oximeter board and the microcontroller. HCS12 determines whether the valve opening needs to be adjusted if oxygen saturation is outside of the desired range. Then, it sends a voltage to the proportional valve using PWM's signal to adjust the valve's opening. Parker HF Pro valve controls the flow of gas proportional to input variable voltage and provides oxygen safe. When there's no input voltage the valve is closed and opens completely when the maximum voltage value is supplied. [63]

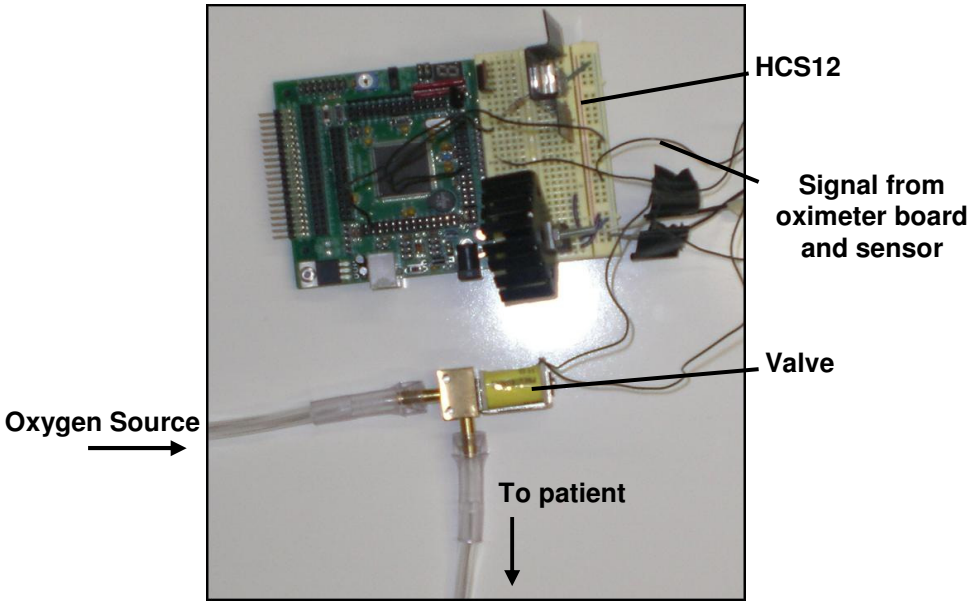


Figure 41: Prototype of the device, showing the Microcontroller HCS12, the proportional valve and also the connections to the oximeter, the oxygen source and the patient. [64]

This is a very recent project from 2008, so the prototype of the device was in implementation at the time of the article's writing. Based on the research made by their developers, the automatic oxygen delivery system showed to be more efficiently than manual delivery systems; the patients did spend effectively more time with optimal oxygen saturation which will effectively decrease the workload of the caregivers with each patient. Furthermore, it is more cost-effective and portable than other devices studied. [63]

4.4. Advantages of the proposed solution

The projected solution is close to the project presented by the Biomedical Engineering Department from the University of Michigan. They have in common some characteristics and goals, as to be safe, efficient and a user-friendly device. However, the system planned has some features that will certainly improve its performance and effectiveness. So, the accelerometer will introduce dynamic parameters in the flow control, making the therapy customized according to each patient's need and degree of motion. The system's optimization is going to pass certainly to the implementation of an oxygen conserving technique, thus allowing the oxygen saving and extending the reservoir duration, improving the patient mobility and reducing the therapy costs. The actuation of the valve using the hardware developed by its manufactures is also an advantage, since it compensates the deviations that occur with temperature rising, thus providing more precise control during its utilization. In addition, the control of the valve is done in current, which is more accurate than in voltage. Furthermore, the proposed device uses a customized valve, according not only with the required flow, but also with the voltage provided by the microcontroller.

5. System Development

5.1. System's Goals

The proposed system intends to adjust the oxygen flow automatically and precisely to a respiratory failure patient based on the SpO_2 readings, measured by pulse oximetry, and movement, measured by accelerometry. The automatic control of the oxygen deliver has two elementary goals: it intends to assure that the patient spends more time with optimal oxygen saturation and realizes the full potential benefits of the therapy and also to prevent the waste of oxygen, thus reducing the total costs of the therapy. As a medical device, it has some other goals as to be safe, rapid to act in response to blood oxygen desaturations and furthermore, it must be easy to use, since it's a device to apply in home oxygen therapy. In addition, it must be lightweight and have small dimensions.

5.2. System Architecture

The system has the architecture shown in the next scheme:

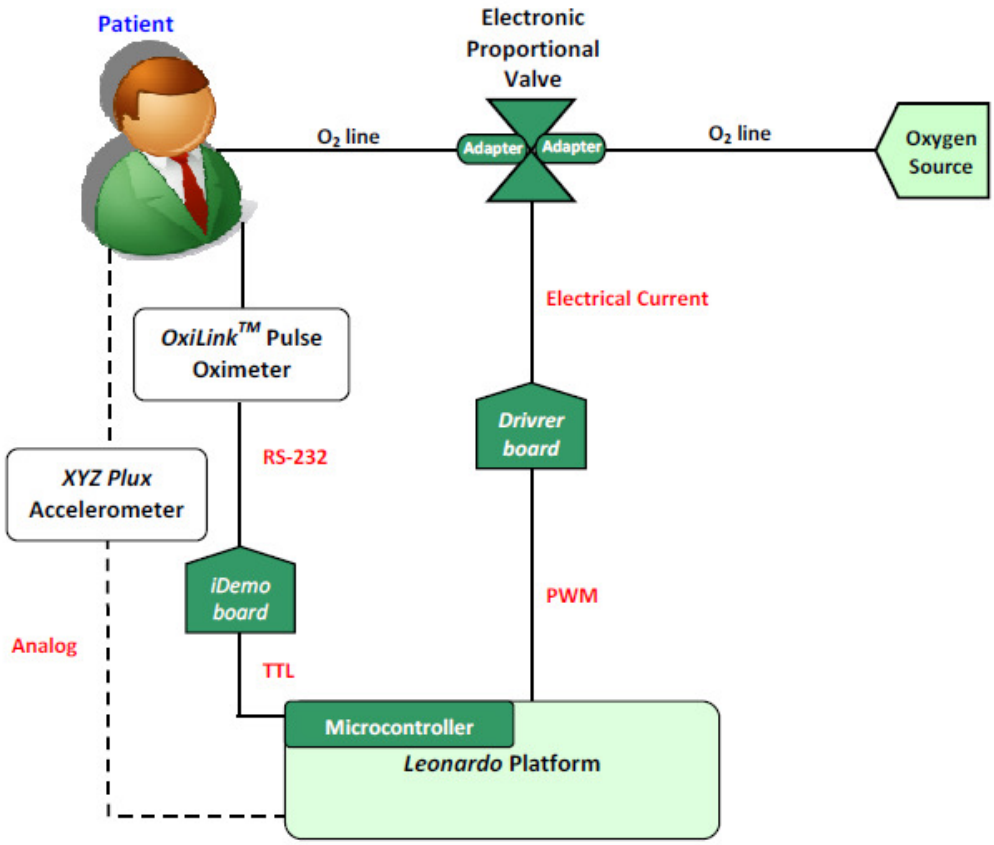


Figure 42: System Architecture, representing the connections between the *Leonardo* Platform, which contains the Microcontroller, the Electronic Proportional Valve, the Driver Board, the OxiLink™ Pulse Oximeter, the iDemo board and the *XYZ Plux* Accelerometer with the patient and the oxygen source. The dotted line corresponds to the part of the prototype not implemented yet.

As represented in the system architecture above, the *Leonardo* Platform, which has integrated the microcontroller PIC24HJ128GP306, controls the all system: it receives through three analogue entries (one for each axis) the data from the accelerometer (XYZ Plux Accelerometer), which gives the information about the patient movements. It also receives the information about the patient's blood oxygen saturation from the pulse oximeter (OxiLink™ Pulse Oximeter), using a finger sensor. The oximeter sends data to the iDemo board through RS-232 and it has a transceiver that converts the input signal to TTL and then sends it to the microcontroller, directly to the U1Rx port (receiving port) of the UART1 module from the microcontroller. Then, the PIC sends a PWM signal to the driver board, using its Output Compare Module in PWM mode, based on a controlling algorithm. The driver board, which is the valve's hardware of actuation, converts the input voltage in a proportional and amplified electrical current to be applied in the electronic proportional valve's solenoid. According to the amount of current, the valve's opening varies, controlling consequently the oxygen delivered from the oxygen source to the patient. Also, some customized adapters are used to safety connect the valve's ports and the oxygen lines.

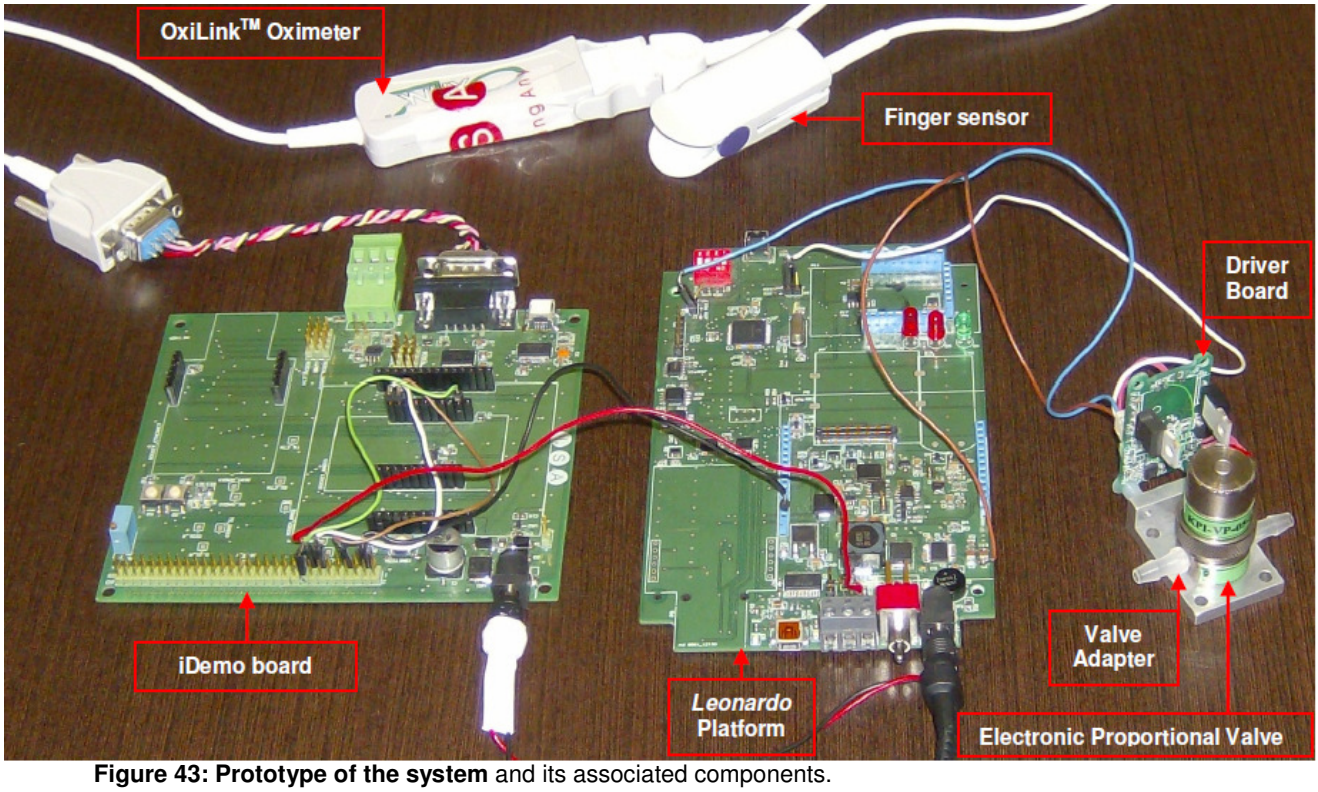


Figure 43: Prototype of the system and its associated components.

The accelerometer isn't included in the prototype implemented for now, although it was planned and studied its incorporation, this will be a future development of the system.

It's therefore a closed-loop system inherently characterized by a close cooperation of different devices: sensors to measure the patient's diagnostic parameters, a controlling algorithm and a therapeutic device that actuates in the oxygen source. As a clinical device, it has to deal with very complex physiological processes and must react promptly and properly on a diversity of physiological conditions and events.

5.3. System Apparatus

During the first stage of the project, the student was focused in the research for devices to apply in the system, specially the pulse oximeter and the medical oxygen valve. Afterward, with the course of the project, it was also needed to purchase other equipments to optimize the system and face the problems that toke place. As result, there were analyzed numerous equipments from several companies and there were made some contacts with their sales managers, in order to assure that the devices' features were according to the system's goals.

As represented in the system architecture, the apparatus is composed by the *Leonardo* Platform, the *Kelly Pneumatics* electronic proportional valve and its hardware of actuation, the driver board, the adapters, the OxiLink™ pulse oximeter and finger sensor and the XYZ Plux accelerometer. Each of the system's components and their technical features are discussed in detail bellow, including the accelerometer although it wasn't included.

5.3.1. The *Leonardo* Platform

The *Leonardo* Platform is a board developed by ISA to remote vital signs monitoring and telemetry applications.

The Platform has many potentialities and includes several components, as a microcontroller, the PIC24HJ128GP306, which is the most important device from the board in the current system. The PIC receives data from the patient that is being monitored; through the pulse oximeter it receives the oxygen saturation readings and through the accelerometer it receives the information about the movement, shock or vibration. The reception of data takes place using wire connections; the oximeter sends data to the iDemo board through RS-232 connection, which communicates with the Platform, directly into the PIC, in its UART module. The accelerometer communicates with the PIC through three analogue entries. With these data, the PIC microcontroller determines whether the valve's opening needs to be adjusted, based on the controlling algorithm. That is, the PIC processes the oxygen saturation value and determines the adequate output flow, supported by an algorithm founded in clinical conditions. To vary the valve's opening, the PIC uses its PWM mode and varies the output compare register, thus changing the output voltage in the output compare pin, to the driver board. As result, the PIC is responsible for the therapeutic device control. The Platform is supplied with 12 VDC, which also supplies the driver board.

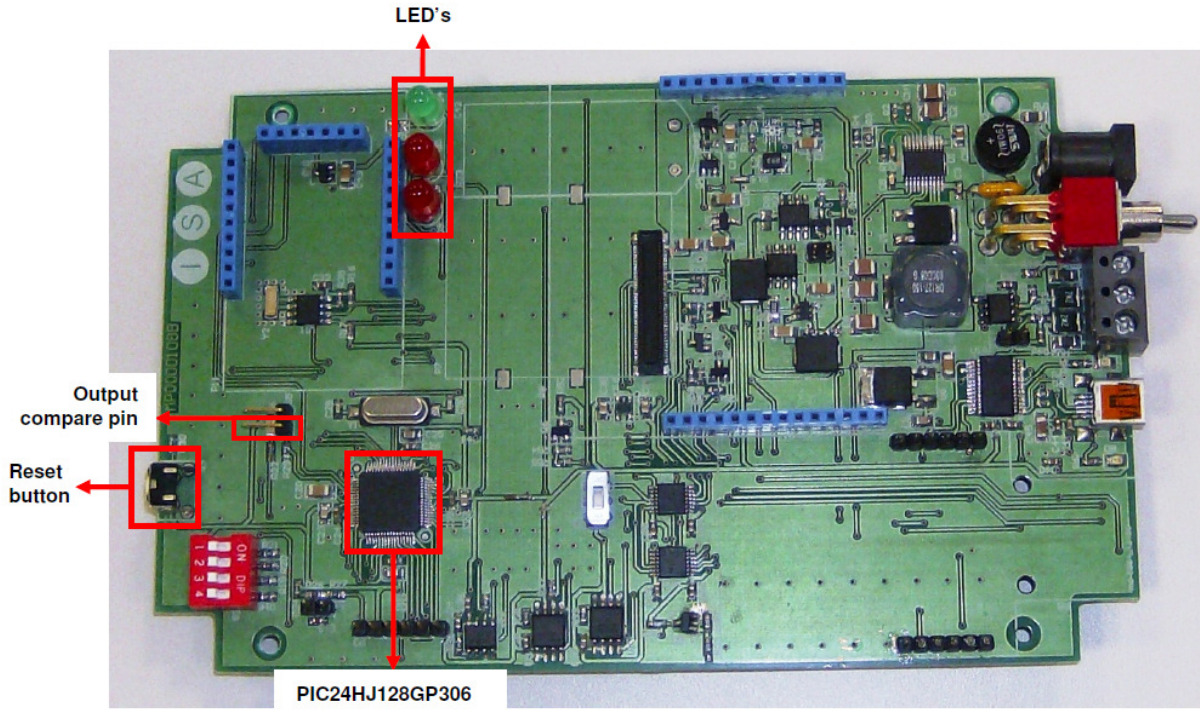


Figure 44: *Leonardo* Platform, showing the microcontroller, the LED's, the reset button and the pin of the output compare module.

The board has a reset button that was used to implement a *panic button*. If the patient, for any reason, fells that isn't receiving enough oxygen or if happens any problem with the valve's opening, the button is pressed and the valve is opened at its maximum diameter, delivering the maximum output flow, which is 7 lpm. When pressed, the button modifies the state of the PIC's pin (from 0 to 3.3V), which is detected by the algorithm and then actuates on the valve opening at its maximum diameter.

The Platform has three LED's, two red LED's and one green LED. The green LED is light when the Platform is being supplied. The red LED's were used to do, not only tests in the system, but also to confirm that the oximeter's data was correctly received, since when the oximeter sensor detects the absence of the finger, the red LED lights.

5.3.2. The Electronic Proportional Valve and the Driver Board

As mentioned before, in the first phase of the project, it was made a research on medical oxygen valves. Those ones were only mechanically and they didn't allow a wide variety of output diameters, additionally, the flow provided was only regulated manually. However, electronic proportional valves, which are used throughout the industry in many flow and pressure control applications, were studied and considered the best option to apply in a device like this. They are small, low powered and can be controlled by voltage/current in a wide variety of output diameters thus allowing a dynamic control of the output flow. Consequently, after some researches, there were selected a few companies that have on the market this kind of equipments. After analysing several possibilities, the valve selected to this project was the miniature proportional valve from *Kelly Pneumatics*. This valve satisfies the project's requirements, since it is low power with small dimensions and has already been used in medical oxygen applications, although not in home care therapy. The valve is normally-closed type with two ports, one inlet and one outlet port. It has a frictionless performance and longevity, tested over 100 million cycles. It was also customized by the manufacturer according to the project's requirements (output flow varying from 0 to 7 lpm). With the proportional valve it was also purchased its actuation hardware, the driver board, also from *Kelly Pneumatics*. It is developed according to the valve's features and it's the ideal solution to operate it. The driver board can be controlled by PWM's output voltage from the microcontroller, which varies from 0 to 3.3V. Accordingly, the voltage sent by the PWM's signal is proportionally converted and

amplified to electrical current by the driver board and then is applied in the electronic proportional valve's solenoid.

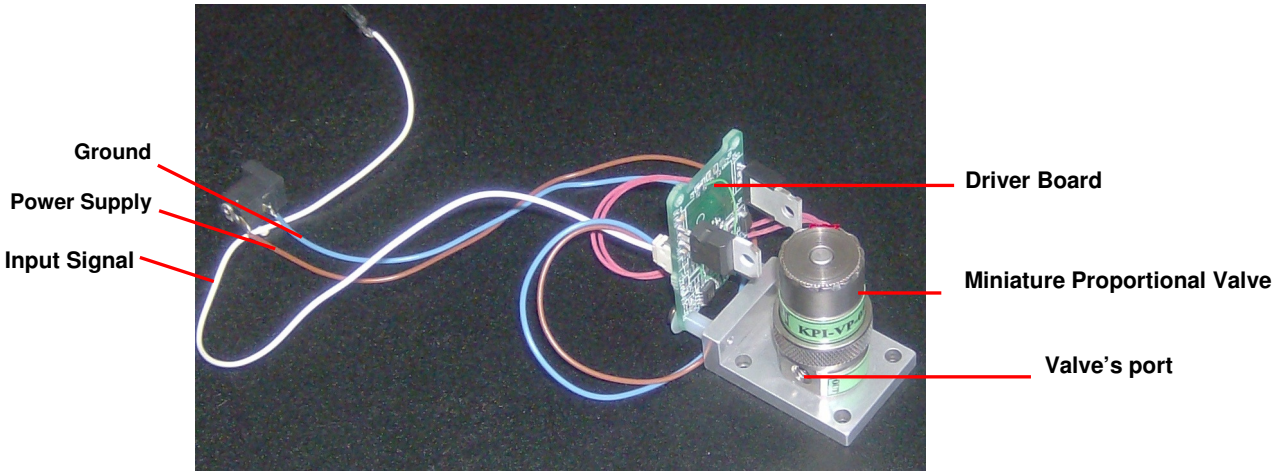


Figure 45: The Miniature Proportional Valve and the Driver Board, from *Kelly Pneumatics*.

Some of the valve and drive board's features are shown in the next table:

Electronic Proportional Valve		Driver Board	
Electrical Connection	Wire leads	**Voltage Required**	12 VDC
Port type	#10-32 Inline	**Operating Temperature**	0-65ºC
Orifice Size	0.010 inch	**Input Signal**	0-3.3 VDC
Maximum Pressure	25 psig	**Power Supply**	<3 W
Voltage Supply	0 to 5 VDC		
Valve Type	2- way		
Operating Temperature	0-50ºC		
Response Time	3 to 5 ms		
Seal Material	Buna-N		
Weight	0.125 lbs		
Output Flow	0 to7 lpm		

Table 6: Technical specifications of the electronic proportional valve and driver board, from *Kelly Pneumatics.*

The working pressure of the valve is according with the source's pressures, since the hospital's sources have 1.5-2 bar (21.75-29 psi) of output pressures.

The flow range of biological significance was considered 0 to7 lpm according to Doctor Moita's guidance, once there aren't available devices with higher flow rates than 5 lpm and it isn't regular practice to deliver more than 5 lpm. So, in preventing unusual cases, the system can provide till 7 lpm.

5.3.3. Valve Adapters

To strongly connect the valve and the oxygen lines adapters are used. They fit in the valve inlet and outlet ports and also in the oxygen tubes. The adapters were provided by Kent Company, who usually provides these equipments to the Kelly's valves. In fact, this company was directly recommended by *Kelly Pneumatics*.

The adapters are straight connectors, threads to flexible tubing and are made of *Kynar* (PVDC-polyvinylidene fluoride), besides they're very cost effective. They are according to the valve's ports dimensions (#10-32 tapered valves) and also with the oxygen cannula inlet dimensions (4 mm) and material (soft materials, as silicone, latex and polyethylene). The coupling enables the safe and easy connection and disconnection of the tubes and fits in the valve's inlet and outlet ports to ensure the correct insulation when oxygen is passing through the lines.

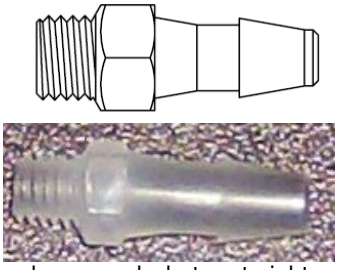


Figure 46: Valve Adapter from Kent, scheme and photo, straight connector thread to 4 mm of inside diameter to #10-32 tapered valves.

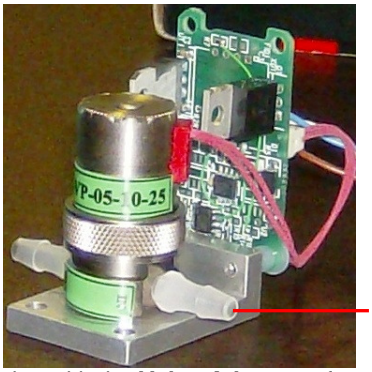


Figure 47: Electronic Proportional Valve with the **Valve Adapters**, from *Kent*, in the inlet and outlet ports.

5.3.4. The Pulse Oximeter

Arterial blood gases must be repeatedly monitored in a non-invasive way, so the integration of a pulse oximeter is very helpful. Its data allows a fast and easy way to access the efficiency of the treatment, because SpO_2<90% are undesirable and negative to the patient's health. The data obtained with the pulse oximeter is used as a feedback signal to maintain the patient SpO_2 within the ideal range, pre-defined by the clinicians. So, the oxygen delivery is increased or decreased based on the data acquired on the patient in the real time monitoring.

After taking into consideration several devices, the OEM pulse oximeter chose to be applied in the system is the OxiLink™ OEM Pulse Oximeter from Smiths Medical. It was taking in consideration many aspects, such as it connector option that should be RS-232, as well as the ability of linking different types of sensors (finger, ear) and also the providence of the communication protocol, which enables the use of the oximeter in integrated solutions.

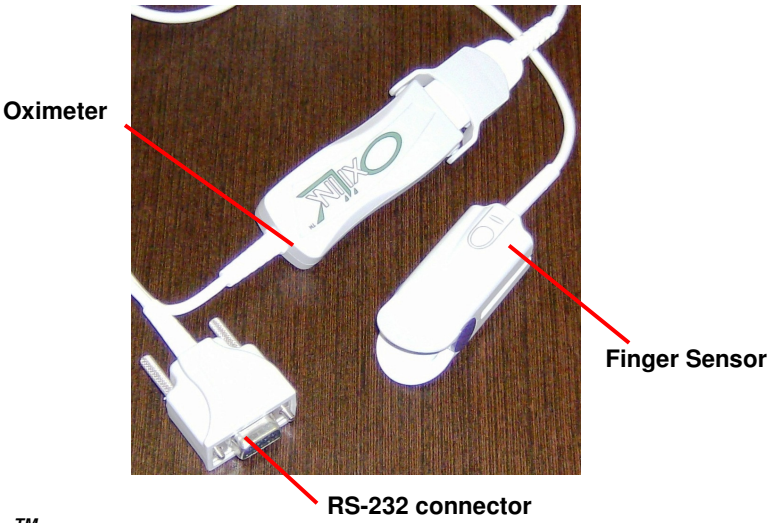


Figure 48: *OxiLink™* Oximeter, from Smiths Medical, showing the finger sensor and the RS-232 connector. The oximeter is contained in the cable.

The OxiLink™ OEM Pulse Oximeter has great flexibility, since it allows the choice of the sensor (finger, ear) and also the connection option, as it has two connector options, RS-232 and USB. Furthermore, it's low power (25 mW), which is important in the system's autonomy and portability.

Since there were some problems with the communication between the oximeter and the *Leonardo* Platform, there was used an iDemo board to establish the communication between both of the equipments. The iDemo board has a transceiver that converts the input signal in RS-232 form to TTL form, which is then received by the PIC's UART module, via UART1 Rx port. Therefore, the OxiLink™ measures the SpO_2

and it's connected to an iDemo board through RS-232 and the iDemo board communicates with the *Leonardo* Platform directly in the UART receiving port.

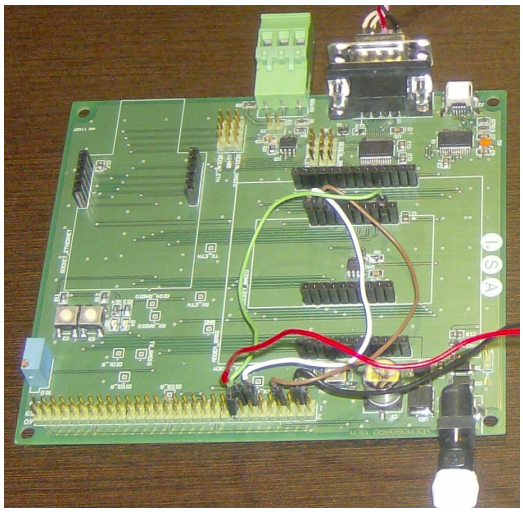


Figure 49: The iDemo board.

5.3.5. Accelerometer

The accelerometer chose to be applied in the system was the *XYZ* Plux accelerometer, from Plux Company. This device is wire connected to the Platform through three analogue entries and incorporates the ADXL330 accelerometer from *Analogue Devices*. It's a three measurement axis accelerometer, with small dimensions (4mm x 4mm x 1.45 mm) and low power (usually 200µA), which measures the static acceleration of gravity, as well as the dynamic acceleration (motion, shock and vibration).

Figure 50: *XYZ* triaxial Plux accelerometer, from Plux.

By applying an accelerometer in the system, it's expected to be possible the control of the oxygen flow according to the patient movements, establishing a dynamic therapy, based on his real and own needs. As well, because each patient is unique, the data obtained with the accelerometer must be considered individually in cooperation and with the support of the clinicians. So, in the first stage, the accelerometer will only be used to acquire data, while the flow rate is controlled only through the oximeter readings. The information collected will be therefore, not only the patient's movements, but also the desaturations that occur due to the daily physical activities. Following, in

cooperation of the clinicians, the data collected during several periods of time will be analysed to establish the pattern to each case.

The final goal is to relate the degree of movement with the saturations/ desaturations to a well known flow, and predict the quantity of oxygen that the patient will need with that degree of motion, avoiding desaturation periods and their negative consequences, especially to the lungs. Like this, the therapy will match more efficiently each patient requirements and it will be according to the mobility of the patient. In fact, with the application of the accelerometer it's expected to increase the patients' mobility and the level of activity, which is highly recommended in pulmonary diseases to slow the illness's progression.

As mentioned before, the accelerometer's integration was studied, but it wasn't performed yet.

5.4. System Features

As it intends to be a safe and user friendly device, the system has to have some features that must be assured. As it must be safe, it has implemented the *panic button*, to assure the continuous flow if some problem happens. In the absence of the finger the system provides a continuous flow of oxygen (2 lpm) and that also increases the device's safety. In addition, the use of the adapters also contributes to the system's security, as they enable the correct connection of the oxygen lines and assure the appropriate insulation.

The device also has to be easy to use; the patient should just put the oximeter sensor in the finger and the oxygen cannula to receive the correct oxygen dose during the daily activities and also the sleeping hours. The usability of the device is therefore assured, since the system is easy to use at home by the patient. Since it is a system to attach in an oxygen source, it should be robust and have small dimensions.

The system must be reliable to promptly react to the desaturations increasing the flow, but also decreasing it when the patient is stabilized, to safe oxygen, extending the container's duration and thus reducing the therapy's costs. As well, the system's response time must be as low as possible, specially to prevent desaturations during higher levels of motion, where the blood oxygen saturation falls rapidly. The reliability of the system depends on the correct communication between the sensors, the *Leonardo* Platform and the proportional valve. Furthermore, the system's performance depends on the correct data acquisition by the sensors and its processing; after that, the system should respond in accordance with the data pre-defined parameters and, as well, it should be able to act effectively on the valve so that the flow of oxygen would be carried out correctly.

It is a customized system, since the valve's actuation is only based in the patient's oxygen saturation monitoring and later in his movement, through a personalized therapy "designed" by the clinicians.

Since the system has small dimensions and is also lightweight, it can be converted in a portable device in the future, once the *Leonardo* Platform has the ability to be supplied with batteries. As mentioned before, there are available three oxygen sources. This system was designed to be applied in a general oxygen source, but the ideal solution would be a portable one, like the LOX portable container's bag, to extend its durability and also to react to the dynamic requests of the patients during their daily activities outside home.

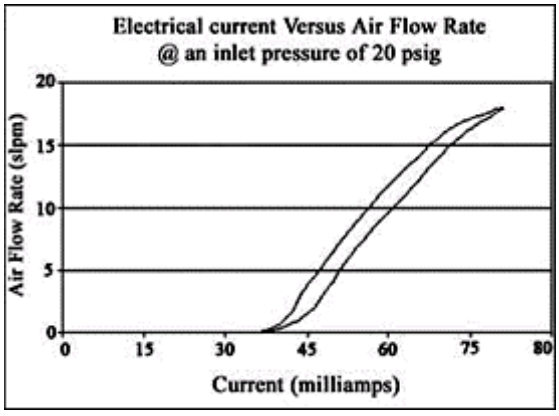
5.5. Other System's Apparatus

Other components, besides those one shown in the logic architecture, were studied to be included in the present device to add functionalities to it. Besides the control of the output flow using the saturation measurements to vary the valve's port diameter, the system also has to have some other apparatus to improve its performance.

The device must allow the oxygen's saving, with the aim of increasing the durability of the source. The implementation of a conservation technique was studied, since it would not only increase largely the source's durability but also contribute to the system's optimization. Currently, there is available a large number of oxygen's conserving valves, which only deliver oxygen during the inhalation. The first approach was to connect the two valves, the oxygen conserver valve and the electronic proportional valve, in series. But, since each one has its associated electronics, both of them could come into conflict and their actuation could be compromised, which would by risky to the patient. So, this approach was putt aside. A second approach studied, but also not implemented, was the modulation of the respiratory cycle using a respiratory effort band. Those bands have piezoelectric sensors that convert kinetic energy into electrical current, enabling the register of the changes in thoracic and abdominal volume due to the expansion of chest and abdomen in the respiratory cycle. The band would therefore detect the inhalation/exhalation and that would enable the modulation of respiratory cycle. If the system could detect the respiratory cycle phase, it would be able to delivery oxygen only in the inhalation. Nevertheless, these hypothesis were only studied and not implemented in practice, so the system hasn't implemented none oxygen's conserving device.

5.6. Valve's Integration and Actuation

The electronic proportional valve is controlled through the electrical current that passes in the valve's solenoid. As it varies, the outlet diameter varies, due to the magnetic field changes, varying the output flow, as well. However, the valve has two typical behaviour curves of electrical current versus flow rate for each working pressure. There are two curves because of the equipment's hysteresis; one for the increasing and another one for the decreasing of the current. This happens because the valve has different behaviours when the electrical current is increased and decreased. This means that it is necessary to know the path that the input followed before it reached its current value when the valve is being actuated. The curves for a standard miniature proportional valve from *Kelly Pneumatics* are presented next:



Graphic 1: Valve's curves of electrical current versus flow rate to an inlet pressure of 20 psig.

The valve can be operated either in electrical current or in voltage. Nevertheless, the control of the valve in current is more effective, so, the driver board operates the valve applying it the desired current. The valve's control in current is more accurate than in voltage, due to the *Joule's Effect*. As the temperature increases, the coil resistance goes down and requires more voltage to maintain the same current. So, the driver board actuates to compensate this effect: it automatically compensates the temperature rising by controlling current to the coil. The flow is proportional to current; therefore, when temperature rises the driver board continues to deliver the same current to the coil thus maintaining the flow steady, regardless of whether the temperature is rising or dropping. Furthermore, the driver board not only compensates the deviations due to the temperature rising, but also the deviations caused by hysteresis, referred previously, which simplifies the valve's control.

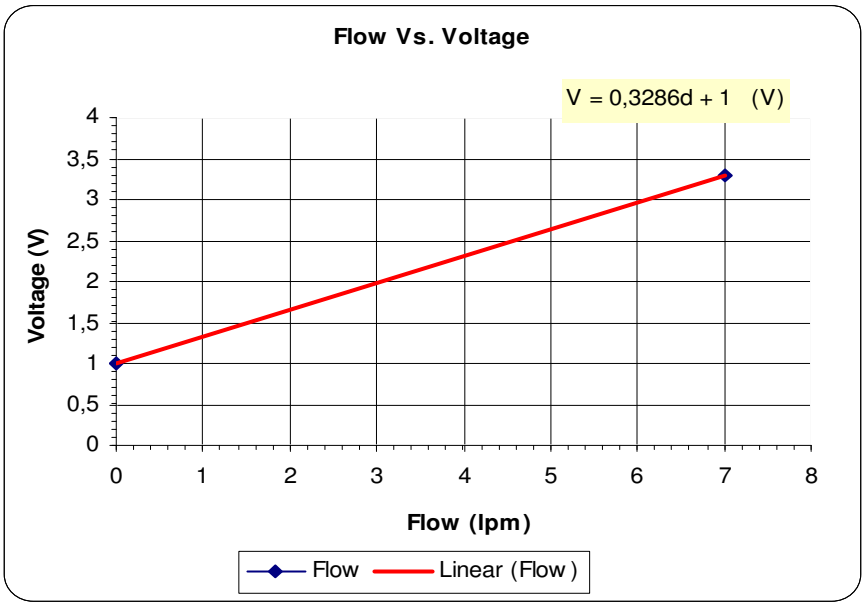
5.6.1. Electronic Proportional Valve Calibration Curve

The PIC actuates the driver board through the PWM's signal, which is a voltage signal. Then, the voltage applied in the driver board is proportional amplified and converted to an electrical current and it is applied in the valve. Therefore, the control of the valve by the driver board is more effective, since it is done in current, and also more precisely, since the driver board is designed to compensate the deviations caused by hysteresis and temperature rising. As mentioned before, the valve was calibrated according to the project features and the manufacturer provided three calibration points, which are indicated in the next table:

Flow (lpm)	Voltage (V)
0	1
7	3.3
12.5	5

Table 7: Valve's calibration points, provided by its manufacturer.

Based on these points, it was obtained the curve that relates the flow with the applied voltage. It was only used two points ((0, 1) and (7, 3.3)), since the useful flow ranges from 0 to 7 lpm and the third point would cause deviations. The referred curve is presented next, as well as the linear fitting:



Graphic 2: Curve of flow rate vs. input voltage based on the flow calibration points.

$$V = 0.3286 \times d + 1 \ (V)$$

Equation 21: Relation between the input voltage and the output flow of the miniature proportional valve.

With this linear fitting, it has been obtained a direct relation between the input voltage and the output flow that enables the development of an algorithm to control the valve using the microcontroller's signals (PWM's signal).

5.6.2. Valve's Control Process

The valve's control is operated through the variations in the applied voltage, which are responsible for the valve's output diameter variations, thus leading to different flow rates. The algorithm receives the value of the oxygen flow rate needed by the patient, d(n), which is determined according to the SpO_2 readings from the pulse oximeter. Using d(n), the algorithm calculates the corresponding voltage, based on the curves' linear fitting obtained above. This voltage is used to determine the percentage of the wave's duty cycle, related to the 3.3V of output voltage supplied by the PIC. It then calculates the secondary output compare register (OCxRS) of the PIC's Output Compare module, which enables to obtain the output voltage required by the PWM mode.

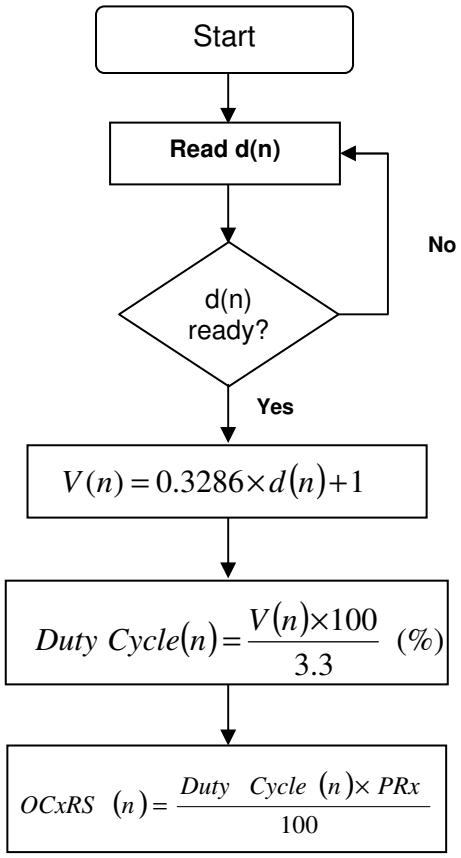


Figure 51: Process's flowchart: with the needed flow rate d(n) is calculated the voltage, using the valve's linear fitting. Then, it's calculated the wave's duty cycle percentage and the PWM's register, OCxRS, to obtain the required output voltage.

This control process is just a voltage control once the electrical current control is entirely made by the driver board based in the input voltage.

The firmware to control and actuate the valve is therefore simple, since it is based in a linear fitting from the valve's calibration points and the driver board prevents the common deviations. Without the driver board, the valve's actuation would certainly be more difficult and it could be less effective and accurate.

To design this control method and operate the valve correctly, the PWM module and the associated timer were previously configured.

The output compare pin used was the OC5 pin from the PIC, which can be view in the *Leonardo* Platform image above. This pin was configured as an output port, through the data direction register (TRIS) of the I/O ports. The TRIS register control bit determines whether the pin is an input or an output; if the TRIS bit is a '1', the pin is an input, while if the TRIS bit is a '0', the pin is configured as an output. The driver board was connected to this pin to receive the input signal.

The PWM module chose was the PWM mode without fault protection, which is used to generate variable duty cycle outputs. That mode was selected through the output compare mode select (OCM) bits in the output compare control register (OC5CON). This mode generates several output pulses due to the changes in the state of the output pin, which is either high or low. The PWM duty cycle is specified through secondary output compare (OC5RS) register, while the output compare (OC5R) register is a read-only compare register.

The OC5RS is continuously compared with the timer value, so, this mode needs a timer to operate. The output compare module works either with the Timer2 or Timer3. The timer used was the Timer2, which was selected by configuring the output compare timer select (OCTSEL) bit in the output compare control register. The timer is clocked with the internal clock source ($F_{OSC}/2$) that was selected through the timer clock source select (TCS) bit in the timer2 control register (T2CON). Furthermore, the prescale factor used was 1:1, which is, each machine cycle corresponds to an increase of timer. That prescale value was configured in the timer input clock prescale select (TCKPS) bit also in timer2 control register. The timer was operated in timer mode by configuring the timer gated time accumulation enable (TGATE) bit in timer2 control register.

The timer2 period register (PR2) enables the selection of the period value of the associated timer. As the PWM frequency used was 3 kHz, the PWM period is 3.33×10^{-4} s and the timer period is obtained using the follow equation:

$$PWM\ \ Period = (PRy + 1) \times TCy \times (TMRy\ \ \text{Prescale Value})$$

The prescale value is 1 and, as the device operating frequency (FCy) is 39628800 Hz, the TCy value is 2.52×10^{-8} s. Using the equation above, the PR2 obtained was approximately 13209.

6. PID Controller's Implementation

6.1. The PID Controller Functionality

The developed algorithm regulates the oxygen flow rate base in an empiric method, constructed according to Doctor Moita's guidance. It was delineated by the clinicians of CHC that, for each 2% of oxygen saturation decreasing, the flow should be increased by ½ lpm. After the stabilization of the blood oxygen saturation, the output flow would be then reduced slowly. Based in these statements, the entire algorithm was developed.

Although, since the system is closed-loop with feedback control, the implementation of a PID controller would be the desirable option to accurately control the output. It would consequently improve the system's performance in the maintenance of the arterial blood oxygen saturation in the ideal range. The PID controller would calculate the input error between the measured oxygen saturation and the setpoint, which should be 90-91% of oxygen saturation. Based on this error, it would calculate the correspondent flow, thus approaching and maintaining the SpO_2 to the setpoint or to an ideal range pre-defined by the caregivers. Hence, the PID manipulates the output in the direction that should move the process toward the setpoint using the automatic mode.

The PID controllers are used in a wide variety of applications, so its implementation involves the calculation of the K_p, K_i and K_d parameters, which must be according to the system that is being controlled. As well, the PID algorithm must be "tuned" for the particular process loop, which is based on the dynamics of the process response, because without such tuning, it will not be able to function correctly.

6.2. PID Computer Implementation

The PID controller's digital implementation in computer consists in the discretization of the integral and derivative presented in the algorithm's equation.

$$u(t) = K_P e(t) + K_I \int_0^t e(\tau) d(\tau) + K_D \frac{de(t)}{dt}$$

Thus, dt is discretized to an interval of time, Δt, between the instants *n* and *n-1*:

$$\Delta t = t_n - t_{n-1}$$

The integral discretization is: $\int e(\tau) d\tau \longrightarrow \sum_0^t e(t)\Delta t \longrightarrow \sum_o^i e_i \Delta(t) + e(t) * \Delta t$

The derivative discretization is: $\frac{de(t)}{dt} \longrightarrow \frac{e(t_n) - e(t_{n-1})}{\Delta t}$

An example of the code to implement in computer the algorithm is presented next:

```
previous_error=0;                    // 1st cycle
setpoint=90;                         // ideal SpO2
dt=1/frequency;
Integral=0;

// define previously the PID coefficients (Kp, Ki e Kd) and the frequency
while (1)
{
  (……..)                                          //Reception of SpO2
error = setpoint – SpO2_mensured;

//Calculation of P, I and D terms

Proportional = Kp*error;
Derivative = Kd* (error – previous_error) /dt;
Integral = Ki* (integral + error*dt);
Output = Proportional + Integral + Derivative;   //Output of the controller
previous_error = error;                          // Error to the next cycle
}
```

As can be seen by the code to the computer implementation of the controller, it involves the continuous reception of the saturation and therefore the calculation of the error between the input oxygen saturation and the defined setpoint, as well as the obtaining of the proportional, integral and derivative terms.

6.2.1. PID Parameters

This algorithm had already been tested in healthcare applications with the same goal as in this project, although not in home care, but in mechanical ventilators and neonatal units' ventilation. Furthermore, in literature there are indicated many approaches to coefficients (K_p, K_i and K_d), obtained using several empiric methods that provide adequate control. Some of these parameters are presented in the next table:

Author	K_p	K_i	K_d
Franklin, 1986	1.000	0.100	2.500
Morozoff, 1994	1.750	0.164	0.1094

Table 8: PID gains obtained by Franklin and Morozoff to accurately control the SpO_2. [53]

These coefficients were tested in a computer simulation with MatLab® using the code presented above, but none of them are adequate to the system. Consequently, the parameters didn't lead to good results, or even satisfactory, in the output flow's calculation. The procedure used was, with an input value of oxygen saturation and with a setpoint of 90%, calculate the error and the output flow. With those parameters, the output flows obtained showed to be too large and inappropriate to the model. However, the computer simulation only considered the flow's determination in consideration of the SpO_2 readings.

6.2.2. Analytic Calculation of the PID Coefficients

As the PID gains from the literature weren't adequate and they're also suggested for other controlling situations, it was analytically calculated the PID parameters using the Ziegler-Nichols step response method.

The analytic calculation of the parameters was based on a curve of patient's open-loop step response from the article, "Oximetry Feedback Flow Control Simulation for Oxygen Therapy", referred in the State of the Art. Although the article doesn't mention the values of the PID parameters used in the computer simulation, it has the representative open loop step response due to a step input of oxygen, which is presented next:

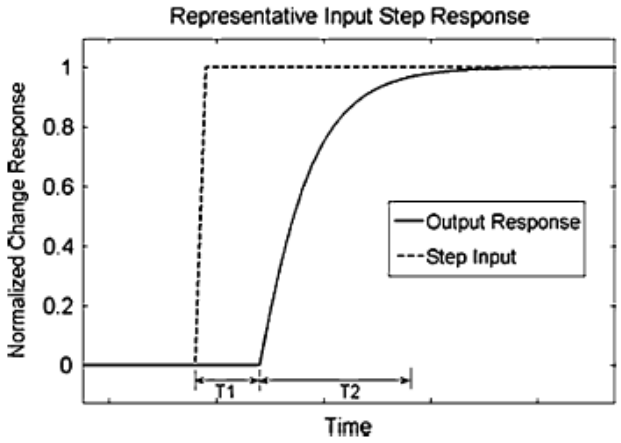


Figure 52: Representative **input step response**, for a step change in the oxygen flow rate. [2]

The article only refers two time parameters, which are the time from onset until a measurable change in saturation, the dead time (T_1), and the lag time (T_2) is the time between the first measurable change and the final stable measurement. Based on literature, the authors considered that T_1 and T_2 are approximately 6 and 15 s, respectively, although these values certainly contain variability between patients and are influenced by numerous aspects. [2]

However, the student faced several problems to calculate the PID parameters, since the patient's mathematical model is unknown, as well as other fundamental parameters; besides the curve of the flow rate step response isn't the best representation, which doesn't enable the closely application of the *Ziegler-Nichols*'s step response method. Nevertheless, using the time constants given by the article and through the application of the equations 11 [65], we have:

Considering two points of non-zero response (t, SpO_2): (6, 0), (21, 1), to draw the response curve:

Slope of curve: $R = \frac{y_0 - y_1}{t_0 - t_1} \Rightarrow R = \frac{0-1}{6-21} \Leftrightarrow R = \frac{1}{15}$

Sampling time: $L = 6s \Rightarrow T_s = 1.8s$

Proportional Gain: $K_P = \frac{1.2}{\frac{1}{15}(6+1.8)} \Leftrightarrow K_P = 2.3077$

Integral Gain: $K_i = \frac{0.6*1.8}{\frac{1}{15}(6+\frac{1.8}{2})^2} \Leftrightarrow K_i = 0.3403$

Derivative Gain:

$$K_D = \frac{0.5}{\frac{1}{15}*1.8} \Leftrightarrow K_D = 4.1667$$

$$K_D = \frac{0.6}{\frac{1}{15}*1.8} \Leftrightarrow K_D = 5$$

Those gains were also tested in a computer simulation with MatLab® using the same code and the same procedure, but they didn't lead to good results. The output flows obtained were also too large and inadequate to LTOT patients.

The output flow values calculated were always compared with the values obtained by the empirical considerations, which were used in the construction of the algorithm.

6.3. Patient's Data Collecting

Seeing that there wasn't obtained good results with the analytic calculation of the PID parameters, it was used a new approach to obtain them. There was collected data, to different values of oxygen flow it was registered the correspondent SpO_2 and also the times, from patients at the Urgency Service, in CHC. There wasn't collected information from the patients of the Pneumology Service since they are already stabilized and their oxygen blood saturation takes much longer to react to oxygen flow fluctuations. Consequently, there were monitored three patients suffering from different pathologies: two men and a woman that were receiving additional oxygen.

The patient 1 was a man with 85 years old with chronic lung edema, who had severe desaturation due to his congested alveoli. The patient 2 was also a man, with 79 years old, with diabetes and obesity. The patient 3 was a woman with 58 years old, suffering from a neoplasia. Although she had acceptable levels of blood oxygen saturation when she was awake, during the sleep she suffered from desaturations and she also had fever, which also caused her saturation's falls. For those reasons, she was receiving oxygen supply.

The oxygen was delivered trough nasal cannulae and the oxygen source used was the ramp medical oxygen from the hospital, which has a flow knob to regulate the quantity delivered. Additionally, the patients SpO_2 measurements were made with each patient pulse oximeter.

The procedure used was the follow: at first, to each patient was cut the oxygen delivery, defining a baseline of 0 lpm and the respective SpO_2. After that, when the SpO_2 stabilized, the flow was incremented to 1 lpm. It was then registered the oxygen's saturation and times to each oxygen's flow, which was varied from 0 to 5-6 lpm. The patient 1, which had severe desaturation was the one with the higher flow rate delivered (from 0 to 6 lpm), the patient 2 hadn't show great variations among the variations of the oxygen flow, so the flow was varied only from 0 to 4 lpm. The patient 3 had the higher baseline to 0 lpm and her oxygen blood saturation became stable only with 1 lpm of flow.

6.3.1. Results

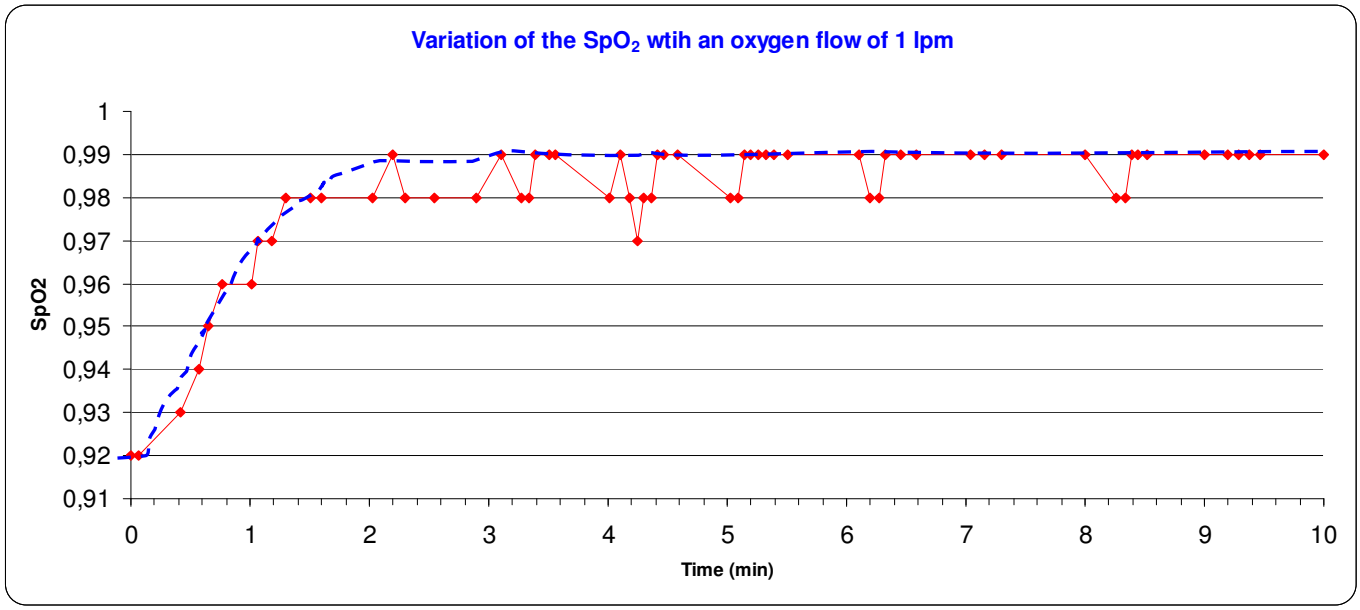
Since each patient suffered from a different pathology, the data obtained has huge variance among each other. Furthermore, the curves obtained aren't according to the expected curve and the deviations between each patient are huge, as well, even the same patient has large variations with different flow rates. Nevertheless, the patient 3 became stable with 1 lpm of flow and her response curve is approximated to an "S", which is the expected variation in oxygen's saturation due to flow's deviation. So, she was chose to the calculation of the parameters. However, in the attachments is presented the collected information from the two other patients (patients 1 and 2), as well as the graphics with their SpO_2 responses due to flow's fluctuations. The data from the patient 3 is presented in the next table:

1 lpm	
Time (min)	**SpO_2**
0	0.92
4/60	0.92
25/60	0.93
34/60	0.94
39/60	0.95
46/60	0.96
1.01	0.96
1.07	0.97
1.18	0.97
1.30	0.98
2.02	0.98
2.20	0.99
2.30	0.98
2.55	0.98
3.10	0.99
3.27	0.98
3.34	0.98
3.39	0.99
3.50	0.99
3.56	0.98
4.01	0.98
4.10	0.99
4.18	0.98
4.25	0.97
4.30	0.98
4.36	0.98
4.42	0.99
4.47	0.99
4.58	0.99
5.03	0.98
5.09	0.98
5.14	0.99
5.20	0.99
5.26	0.99
5.33	0.99
5.39	0.99
5.50	0.99
6.10	0.99
6.20	0.98
6.27	0.98

6.33	0.99
6.45	0.99
6.59	0.99
7.04	0.99
7.15	0.99
7.30	0.99
7.50	0.99
8	0.99
8.26	0.98
8.34	0.98
8.39	0.99
8.44	0.99
8.52	0.99
9	0.99
9.20	0.99
9.28	0.99
9.38	0.99
9.47	0.99
10	0.99

Table 9: Patient 3 data collected at the Urgency Service, with the blood oxygen saturation variations and correspondents times with a flow rate of 1lpm.

With these data it was obtained the graphic that represents the variation of the oxygen saturation with an oxygen flow of 1 lpm. As can be view, the curve obtained isn't exactly the step response in "S", although it can be approach to it. The dotted line represents a simple approach of the curve to an open loop step response of a first order system.



Graphic 3: SpO_2 variations from patient 3, obtained by varying the flow from 0 to 1 lpm. The red line represents the patient response to the flow variation, while the blue dotted line is a simple approach of the curve to an open loop step response of a first order system.

6.3.2. Data Analysis and Calculation of the PID Parameters

From the graphic and the table with the information, the dead time is L=25 s, since the SpO_2 of the patient took 25 s to increase from 92% to 93%. After 5.14 min, the SpO_2 tended to stabilized in 99%, the final stable measurement, although there have occurred a few oscillations to 98%. As well, the patient's SpO_2 took much longer to stabilize than the expected, since in the referred article, the time between the first measurable change and the final stable measurement is 15 s. Consequently, it will severely influence the calculation of the gains.

Sampling time: $L = 25s \Rightarrow T_s = 7.5s$

Considering two points of non-zero response (t, SpO_2): (4, 0.92), (308.4, 0.99)

Slope of curve: $R = \frac{y_0 - y_1}{t_0 - t_1} \Rightarrow R = \frac{0.92 - 0.99}{4 - 308.4} \Leftrightarrow R = 0.0003$

Proportional Gain: $K_P = \frac{1.2}{0.0003(25 + 7.5)} \Leftrightarrow K_P = 123.077$

Integral Gain: $K_i = \frac{0.6 * 25}{0.0003(25 + \frac{7.5}{2})^2} \Leftrightarrow K_i = 60.491$

Derivative Gain:

$$K_D = \frac{0.5}{0.0003 * 7.5} \Leftrightarrow K_D = 222.222$$

$$K_D = \frac{0.6}{0.0003 * 7.5} \Leftrightarrow K_D = 266.667$$

Those PID parameters are much higher than the gains suggested in literature (see table 8) and, as expected, they aren't satisfactory to apply in the model and there weren't good results when simulated in *Matlab*.

6.3.3. Conclusions

One of the major challenges in SpO_2's precise control is the large unknown parameters of the model, as well as the time delays and constants, which differ for each patient. Due to this variability, the model robustly can be seriously affected. With the data collected it has verified that each pathology lead the patients to different responses to the oxygen. Additionally, the SpO_2 had been measured using different pulse oximeters. Thus, different equipments have different sensors, as well as different calibrations, leading to changes in the results between each patient.

Therefore, although it had been obtained the PID parameters, they do not lead to good results in PID computer implementation to control the flow. Additionally, the time constants obtained differ from the expected and also the curve used to calculate the parameters doesn't exactly match with an open loop step response of a first order system, which also introduced errors in the parameters' calculation.

7. Prototype's Tests

7.1. Plan of Tests

The prototype was tested in three different stages as the control algorithm was developed and optimized.

The first stage of tests was made with the software *Terminal* and a red LED to simulate the oximeter's data and the valve's opening, respectively. Since the PIC doesn't vary the output voltage analogically, there was used a LED to do the tests. It was placed on the output compare pin to be actuated by the PWM's signal and *Terminal* transmitted information from the PC to the PIC that received it via UART1 (U1RX pin). Furthermore, it was used an oscilloscope to observe the square wave's width varying with the received information.

The second stage consisted in computer simulations that were made already with the oximeter and the valve, but without the oxygen source. With the real saturation data, it was observed the registers in MPLAB® that indicated the saturation's readings, the correspondents flow and voltage to be applied in the valve.

The last stage of tests was done at the Pneumology Service from CHC and in real situations; this stage had two phases.

In the first phase, the tests were made with a COPD patient that usually do home oxygen therapy. The patient putted the oximeter sensor in his finger and the prototype was connected to the oxygen lines.

It was verified the receiving and delivering of oxygen by the valve's ports and the flow was measured using a hospital's flowmeter, from *Gasin* medical company. This flowmeter is used by *Gasin*'s technicians at the patient's home to support that the person is receiving a certain flow in case of doubt. The flowmeter has a sphere inside a graduated tube that is elevated with the oxygen's flow. Therefore, the quantity of oxygen supplied is read by the sphere's position in the scale of the tube, as can be seen in the figure below:

Figure 53: Flowmeter from *Gasin*, the sphere inside the tube is elevated with the flow provided; the maximum flow read is 8 lpm.

The oxygen source used was the hospital's ramp, which has a flow regulator knob to manually control the output flow rate and it has an adapter to the oxygen lines connection.

Figure 54: Oxygen ramp source from the hospital, with the oxygen line connected in its adapter.

The tests were done with a higher flow than needed; typically 4 lpm of inlet flow, and the output was regulated by the valve outside diameter. These tests enable the detection of the lack of insulation between the valve's ports and the oxygen lines.

In the second phase, there were made tests in the same conditions, using the hospital's oxygen source and the flowmeter from *Gasin*, but with a healthy person. However, the valve had already the correct adapters in its ports, which allowed the insulation and the quick connecting and disconnecting of the tubes. Since the common oxygen cannulae have inside diameters larger than 4 mm, the oxygen lines had to be customized. As the valve's adapters are adequate to oxygen lines with 4 mm of inside diameter, *Gasin* provided an oxygen tube with this diameter. Nevertheless, the inside diameter of the flowmeter was the same of the previous tests, thus leading to small wastages in the outlet port of the valve, although, those wastages were much lesser than in the previous tests, as a result of the adapters presence.

The prototype was tested in different situations; the first test was made with the following procedure: the flow was set higher than needed, 2 lpm, and with the oximeter readings was established the correct output flow through the variation of the valve's diameter. The second test was made in the absence of the finger and the flow was set again higher than needed, 3 lpm. Finally, the last test done was a conjugation of the

two previous tests: the flow was set higher than needed, 4 lpm, and with the oximeter's readings was established the correct flow; then it was absent the finger from the sensor to test the system reaction to that absence.

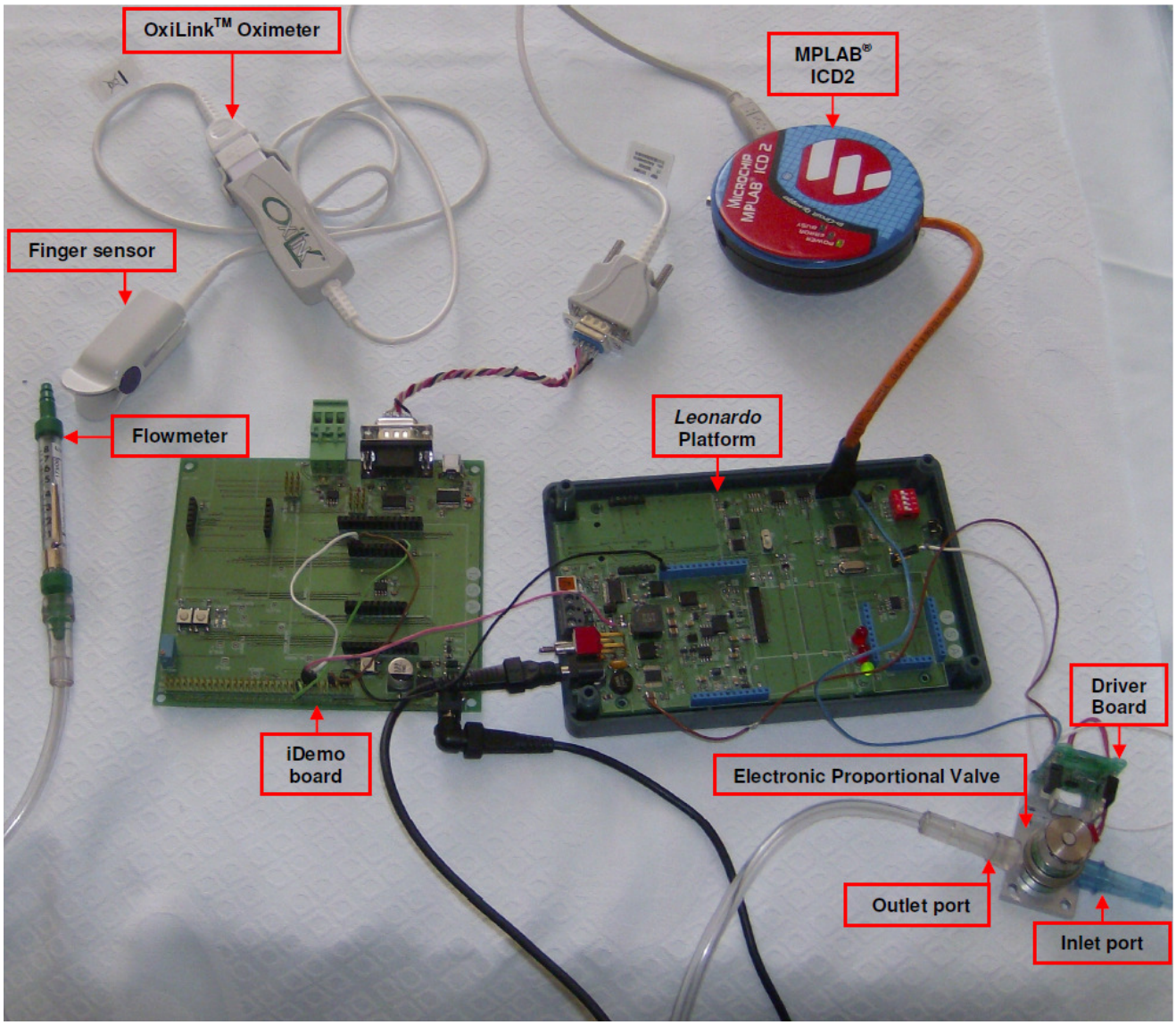


Figure 55: Prototype of the system to automatically regulate the output flow; the valve has connected to its inlet port the oxygen line customized with 4 mm of inside diameter and the outlet port has attached the *Gasin*'s flowmeter.

7.2. Results and Discussion

In the first stage tests, it was confirmed that the LED's brightness changes with different data sent; when the saturation sent was higher the LED's brightness was lower, while when it was lesser the brightness was higher. That is consistent with the expected results, since with higher oxygen saturation is need less oxygen, so the valve needs to be less open and the applied voltage is therefore lesser, with a low duty cycle percentage, as the width of the pulse is lesser. When the oxygen's saturation falls, the valve needs to be more opened, so the applied voltage is higher due to a higher duty cycle percentage and the higher wave's width. The results obtained can be visualized with the follow graphic:

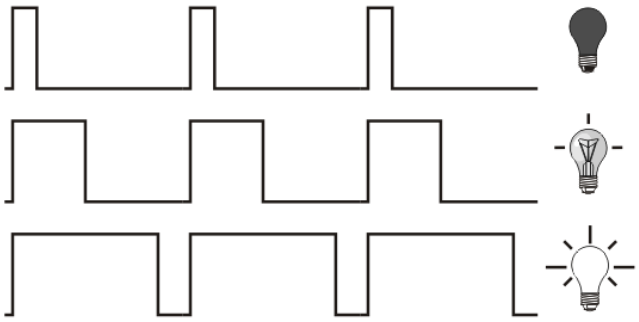


Figure 56: Wave's duty cycle variation tested with a LED, as the wave's duty cycle changes, the LED's brightness changes, because higher widths of the wave are associated with higher voltages, so the LED's brightness is higher, while with lower widths, the brightness is lesser.

The PWM signal has a variable duty cycle percentage because of the changeable width of the pulse and the square waves with different duty cycles (50%, 25%, 90%,…) were visualize with the oscilloscope.

In the second stage of tests, with the real saturation data, it was observed the registers in MPLAB® and they indicated the saturation readings from the oximeter, the correct output flow and also the correspondent voltage to be applied in the valve. The results were also consistent with the expected theoretically, since it could be confirmed that the output voltage was according to the required flow, by means of the valve's calibration curve.

In the first phase of tests at the Pneumology Service, although the valve received and delivered oxygen, the flowmeter wasn't adjust to the valve's port as well as the oxygen cannula, and there were several oxygen's wastages in the oxygen pipes due to the lack of insulation. So, it couldn't be measured the oxygen in the outlet port of

the valve and there wasn't a well known quantity of oxygen in the inlet port. Thus lead to the system's improvement with the integration of the correct valve's adapters.

In the second phase of tests at the Pneumology Service the results obtained were also positive. In the first test, the flow was set higher than needed, 2 lpm, and the person had 97% of SpO_2, whose correct flow would be 0.5 lpm. The valve regulated the output dose and it was confirmed with the flowmeter that the real output flow rate was actually 0.5 lpm. Therefore, with the developed firmware solution, the flow was regulated as it intended to be.

In the second test, the flow was set again higher than needed, 3 lpm, and without the finger in the sensor, the system set the flow in 2 lpm, as expect.

In the last test, the flow was higher than needed, 4 lpm, and with the oximeter's readings of a healthy person it was established the correct output flow, which was 0.5 lpm. Then, it was absent the finger from the sensor. The system promptly react to that absence and not only the LED has lighted, but also the flow was set higher, from 0.5 lpm to 2 lpm.

The lack of insulation between the outlet port and the flowmeter is because of the flowmeter's tube inside diameter; the tube is not fully connected to the adapter, it is adapted to the larger tubes. The flowmeter should also have an inside diameter of 4 mm in its line to be according to the adapter's size, however the common oxygen cannulae have inside diameters of 6 mm, the flowmeter is custom-made to them. So, when the input flow was 2 lpm in the absence of the finger, the output flow should also be 2 lpm. However, due to these small losses in the connection of the flowmeter, the reading was not 2 lpm, but slightly lower. Thus, taking into account the existence of some deviations, the results are in line with the expectations. In fact, the measurement of the flow isn't entirely accurate, being affected not only by the losses in the connecting, but also by the reading of the user, given that the sphere position on the graduated scale isn't unique, the readings can be affected by it.

Those results are according to the wished-for the clinicians and they are also in concordance with the project's goals.

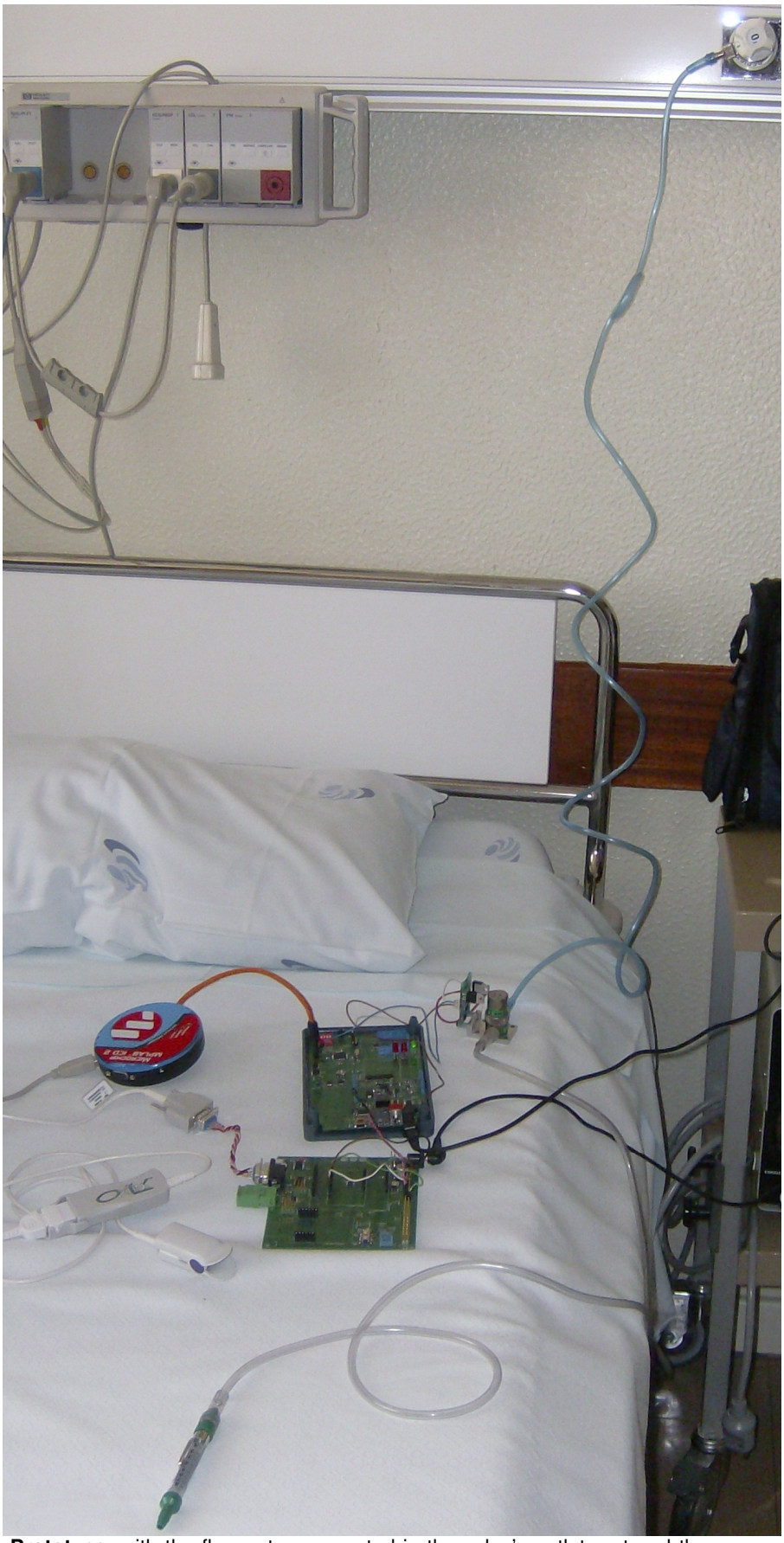


Figure 57: Prototype, with the flowmeter connected in the valve's outlet port and the oxygen line in the inlet port, attached to the ramp source, during the tests at the Pneumology Service, CHC.

8. Future Works

Future works will certainly pass to incorporate the accelerometer to register the daily physical activity of the patient and correlate it with the oxygen desaturations. To the first stage, where the accelerometry data will only be recorded, it will be necessary to use the memory of the PIC, which was not addressed in this project. With the collected data, it will be necessary to optimize the algorithm's response to each patient's therapy.

It also must be implemented an oxygen conserving technique to delivery oxygen only during the inhalation or even only during its early stage, when oxygen is more valuable. The oxygen saving is one of the system's goal and to achieve it, the system must have a conservation technique implemented, with the patient's respiratory cycle preview.

The algorithm developed can also be improved and the implementation of a PID controller would be the ideal in a system like this, since it would lead to a more accurate control. So, the re-calculation of the PID parameters or even a new data collection on patients in a controlled laboratory environment would be the desirable option.

The *Leonardo* Platform isn't the ideal solution to this system, since there were several problems with the reception of the oximeter's data, which was solved using an iDemo board. This isn't the best to the system's portability, so in the future the *Leonardo* Platform can be replaced by other board more appropriated and designed according to the system's features and requirements.

The system can be developed to become portable, through the application of batteries in the *Leonardo* Platform. The Platform has this ability and the battery will supply not only the Platform, but also the driver board, the oximeter and the accelerometer. That is perfectly possible, since the device's components are low powered, as well, to convert the system in a portable device, it should be taken in account the apparatus consumptions to extend the batteries duration.

As the proportional valve is normally-closed type, as a matter of security, the system should also have another valve, parallel to the proportional valve, to be opened manually if the electrical current or the battery fails. That would assure the continuous supply of oxygen and a system such as a tap would be enough.

The prototype should also be tested in several circumstances; not only in static conditions with the patient sited, but also in dynamic conditions, with the patient moving or even running in a crosswalk.

9. Conclusion

Oxygen therapy using a proportional valve will control more accurately the oxygen flow rate to the patient; the valve can be precisely controlled using the patient's data in an easy and non-invasive way. The dynamic control of the oxygen delivery will certainly improve the patient's life quality. The control based on patient own data enables the flow to exactly match the demands that represents a huge step in the LTOT.

The system showed to have fast response to oxygen desaturations, immediately giving the correspondent flow and applying the correct voltage to the driver board. The rapid response is especially important in dynamic conditions, such as movement, where the patient's blood oxygenation levels fall abruptly.

The purpose of saving oxygen was partly achieved, because in cases where the blood oxygen saturation figures above 90%, the rate of oxygen was gradually decreased. Thus, there are substantial savings compared to the continuous and constant flow of 2 lpm, typically used in home-based treatments.

In addition, the use of adapters in the valve's ports increases the system's safety. As well, the system's security is assured by other features; if the patient doesn't have the oximeter correctly positioned in the finger, the device will also continue to supply oxygen in a continuous way, 2 lpm, which will certainly assure the correct levels of the SpO_2. Also, the *panic button* can be pressed in any emergency case.

The clinical foundations of the controller algorithm will certainly enable the patient to spend more time in safe levels of SpO_2, slowing the progress of the disease, as well as extending the tank's duration.

However, besides the positive results, the system needs to be improved in order to assure the patient safety, comfort and life's quality. Nevertheless, the majority of the project's goals were achieved and there were also solved the several problems that the system faced during its implementation.

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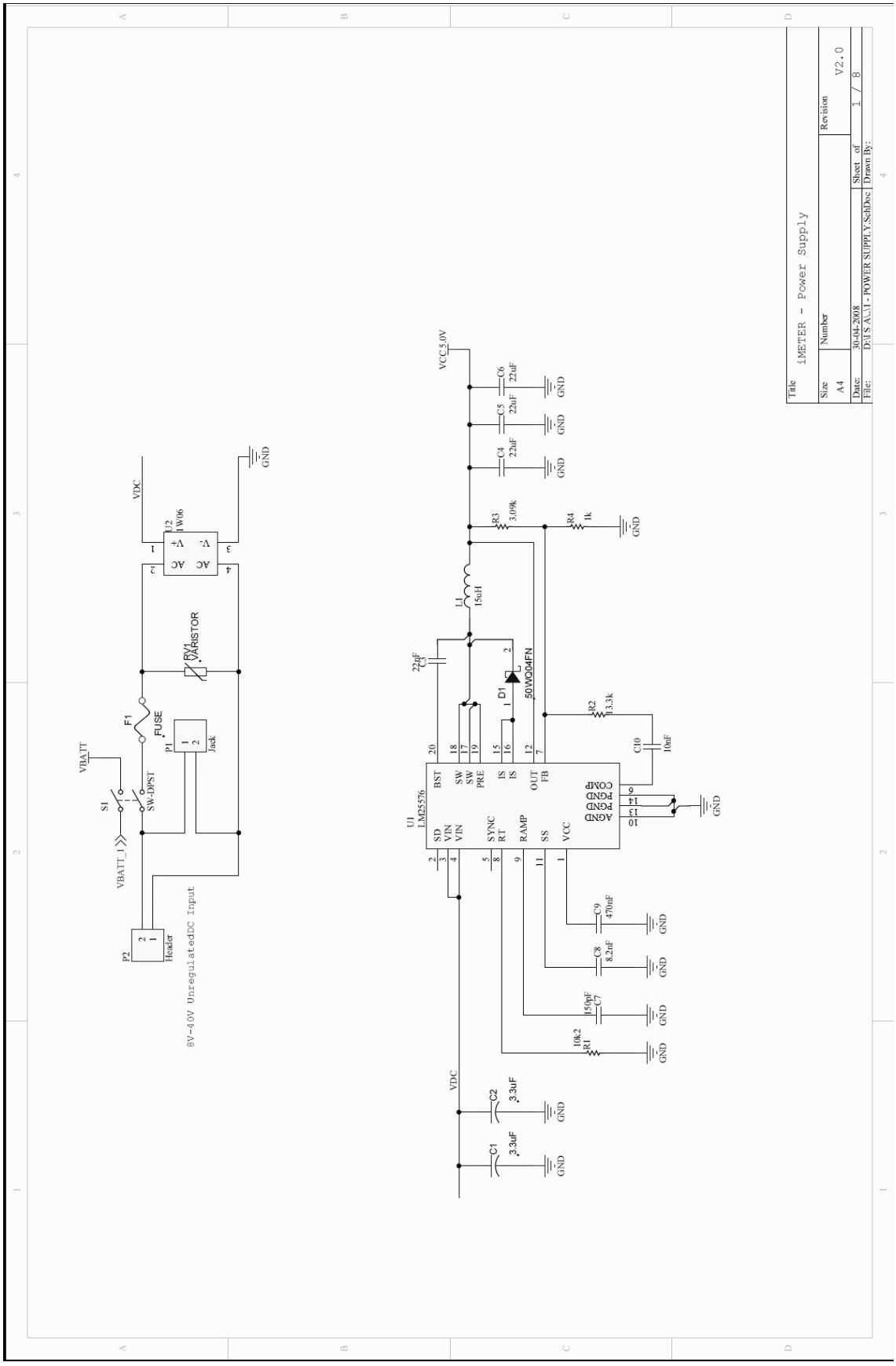
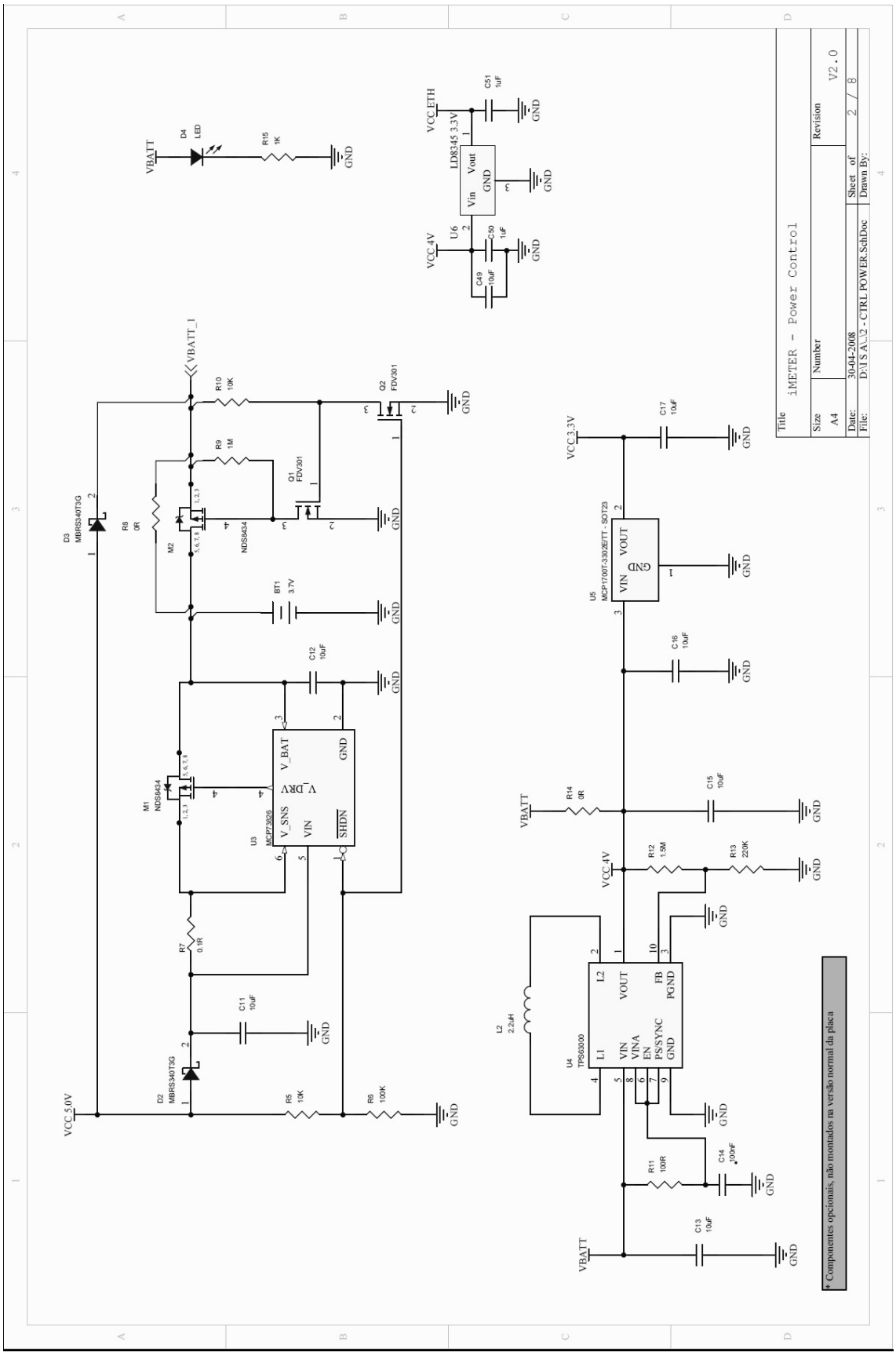
Attachments

Schematics of the *Leonardo* Platform

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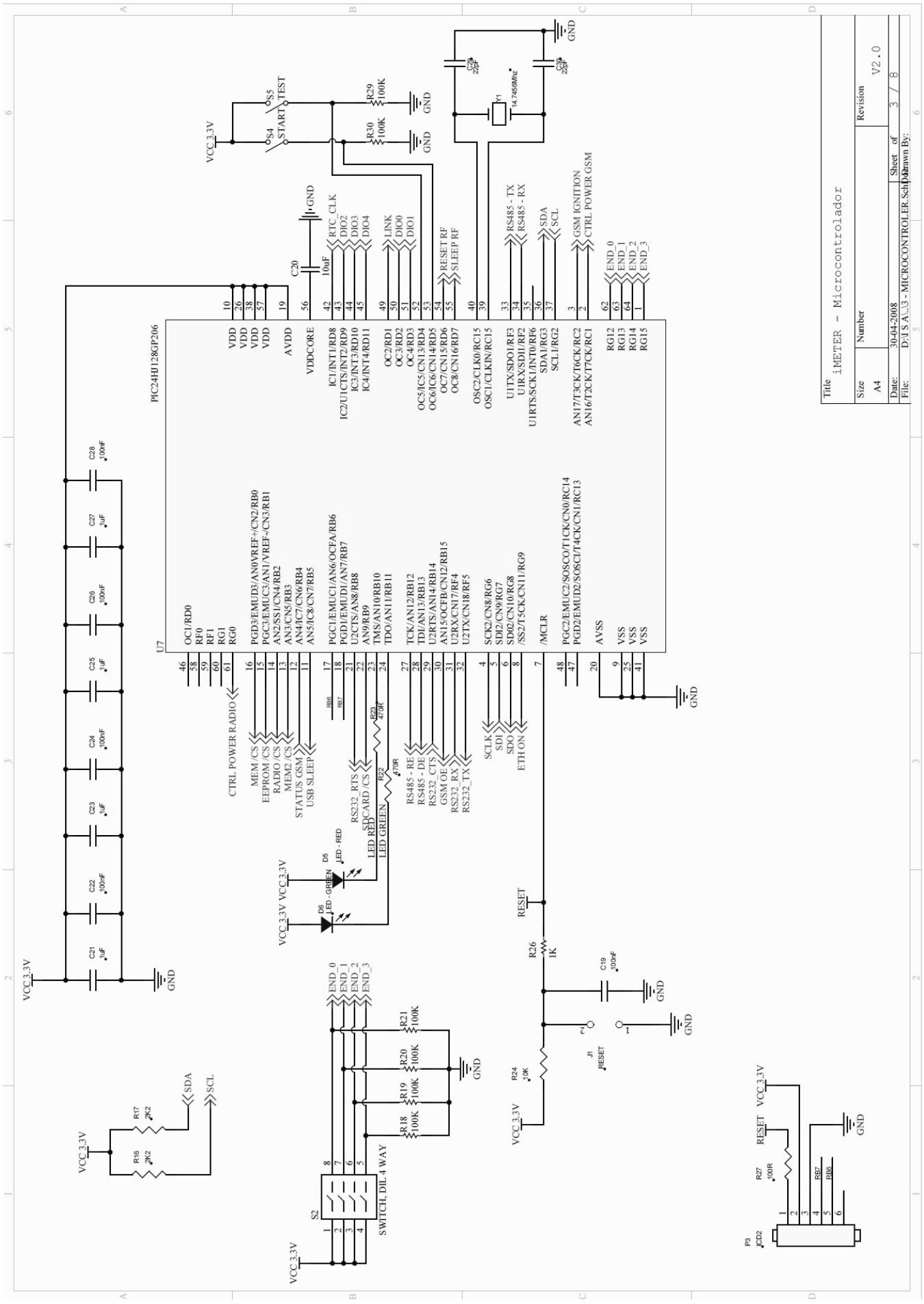
iMETER - Power Control
Size
A4
Number
Revision
V2.0
Date:
30-04-2008
Sheet of
2 / 8
File:
D:\IS A\..\2 - CTRL POWER.SchDoc
Drawn By:
* Componentes opcionais, não montados na versão normal da placa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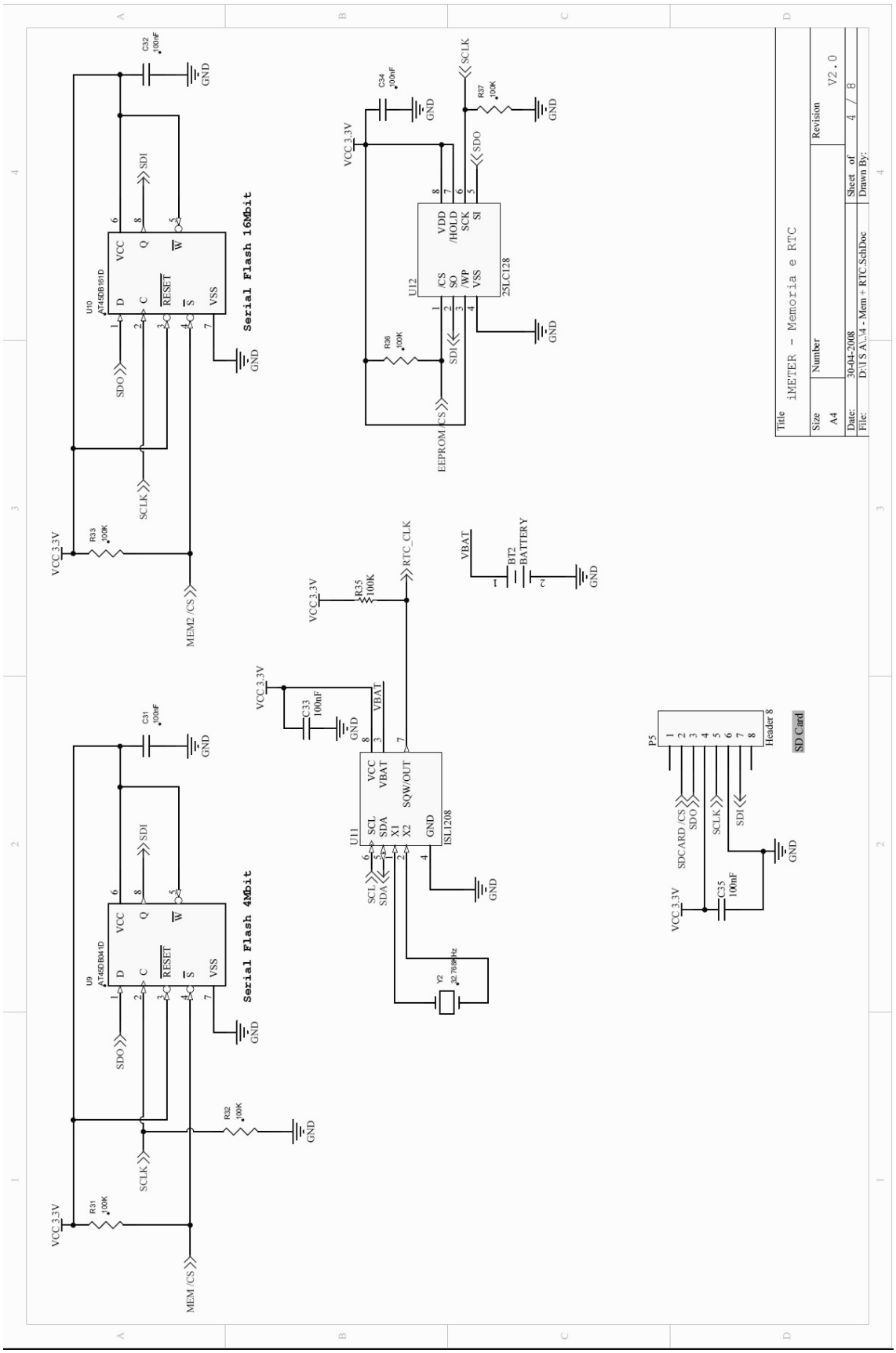
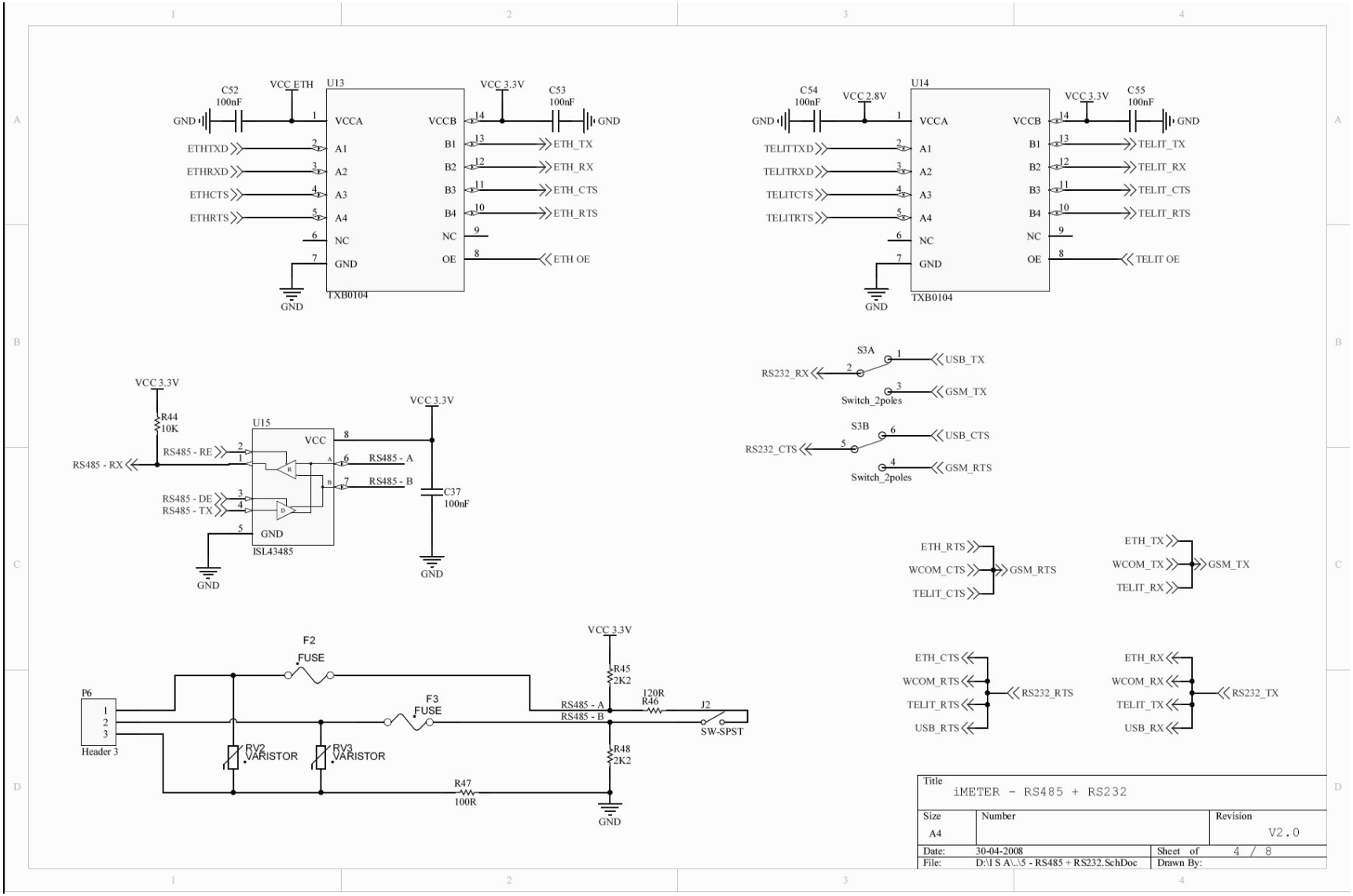
iMETER - Microcontrolador
Size
A4
Number
Revision
V2.0
Date:
30-04-2008
Sheet of
3 / 8
File:
D:\I S A\...\3 - MICROCONTROLER.SchDoc
Drawn By:
PIC24HU128GP206
U7
VCC_3.3V
GND
SDA
SCL
RESET
SWITCH, DIL 4 WAY
S2
S4
START
S5
TEST
Y1
14.7456Mhz
RTC_CLK
LINK
RESET RF
SLEEP RF
RS485 - TX
RS485 - RX
GSM IGNITION
CTRL POWER GSM
LED - RED
LED - GREEN
LED RED
LED GREEN
SDCARD /CS
RS232_RTS
RS485 - RE
RS485 - DE
RS232_CTS
GSM OE
RS232_RX
RS232_TX
SCLK
SDI
SDO
ETH ON
CTRL POWER RADIO
MEM /CS
EEPROM /CS
RADIO /CS
MEM2 /CS
STATUS GSM
USB SLEEP
END_0
END_1
END_2
END_3
JCD2
P3
J1


Serial Flash 16Mbit
Serial Flash 4Mbit
U10
AT45DB161D
U9
AT45DB41D
U12
25LC128
U11
ISL1208
C32 100nF
C31 100nF
C34 100nF
C33 100nF
C35 100nF
R33 100K
R31 100K
R32 100K
R36 100K
R37 100K
R35 100K
B1 2 BATTERY
Y2 32.768kHz
P5 Header 8
SD Card
VCC 3.3V
GND
SDI
SDO
SCLK
RESET
MEM /CS
MEM2 /CS
EEPROM /CS
SDCARD /CS
RTC_CLK
VBAT
SCL
SDA
Title iMETER - Memoria e RTC
Size A4
Number
Revision V2.0
Date: 30-04-2008
Sheet of 4 / 8
File: D:\I S A\...\4 - Mem + RTC.SchDoc
Drawn By:


C52
100nF
VCC ETH
U13
VCC 3.3V
C53
100nF
GND
VCCA
VCCB
ETHTXD
ETHRXD
ETHCTS
ETHRTS
A1
A2
A3
A4
NC
GND
B1
B2
B3
B4
NC
OE
ETH_TX
ETH_RX
ETH_CTS
ETH_RTS
ETH OE
TXB0104
C54
100nF
VCC 2.8V
U14
VCC 3.3V
C55
100nF
TELITTXD
TELITRXD
TELITCTS
TELITRTS
TELIT_TX
TELIT_RX
TELIT_CTS
TELIT_RTS
TELIT OE
S3A
RS232_RX
USB_TX
GSM_TX
Switch_2poles
S3B
RS232_CTS
USB_CTS
GSM_RTS
R44
10K
U15
VCC
RS485 - RX
RS485 - RE
RS485 - DE
RS485 - TX
RS485 - A
RS485 - B
C37
100nF
ISL43485
ETH_RTS
WCOM_CTS
TELIT_CTS
ETH_TX
WCOM_TX
TELIT_RX
ETH_CTS
WCOM_RTS
TELIT_RTS
USB_RTS
RS232_RTS
ETH_RX
WCOM_RX
TELIT_TX
USB_RX
RS232_TX
F2
FUSE
F3
FUSE
P6
Header 3
RV2
VARISTOR
RV3
VARISTOR
R45
2K2
R46
120R
J2
SW-SPST
R48
2K2
R47
100R
Title
iMETER - RS485 + RS232
Size
A4
Number
Revision
V2.0
Date:
30-04-2008
Sheet of
4 / 8
File:
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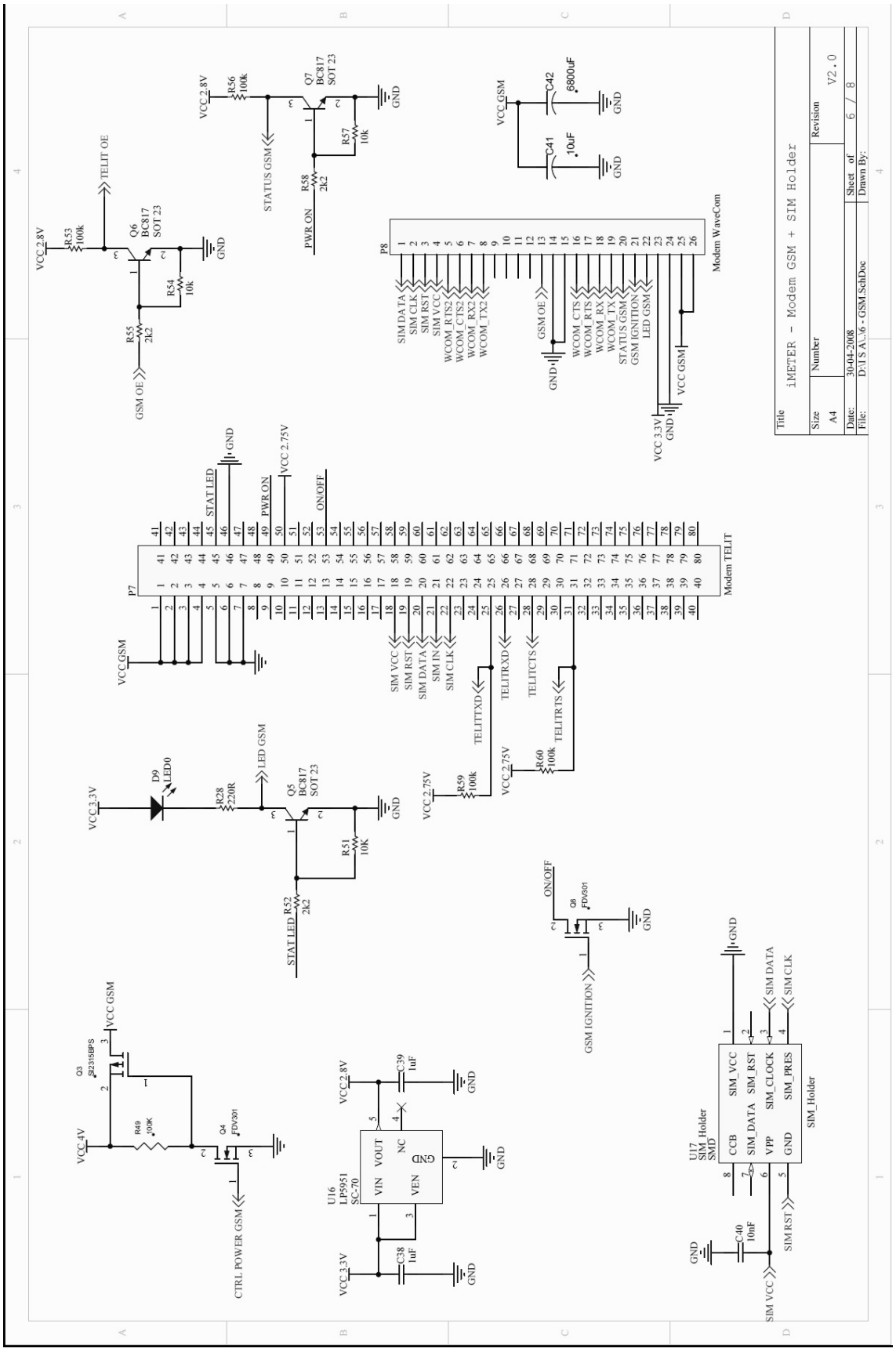
Drawn By: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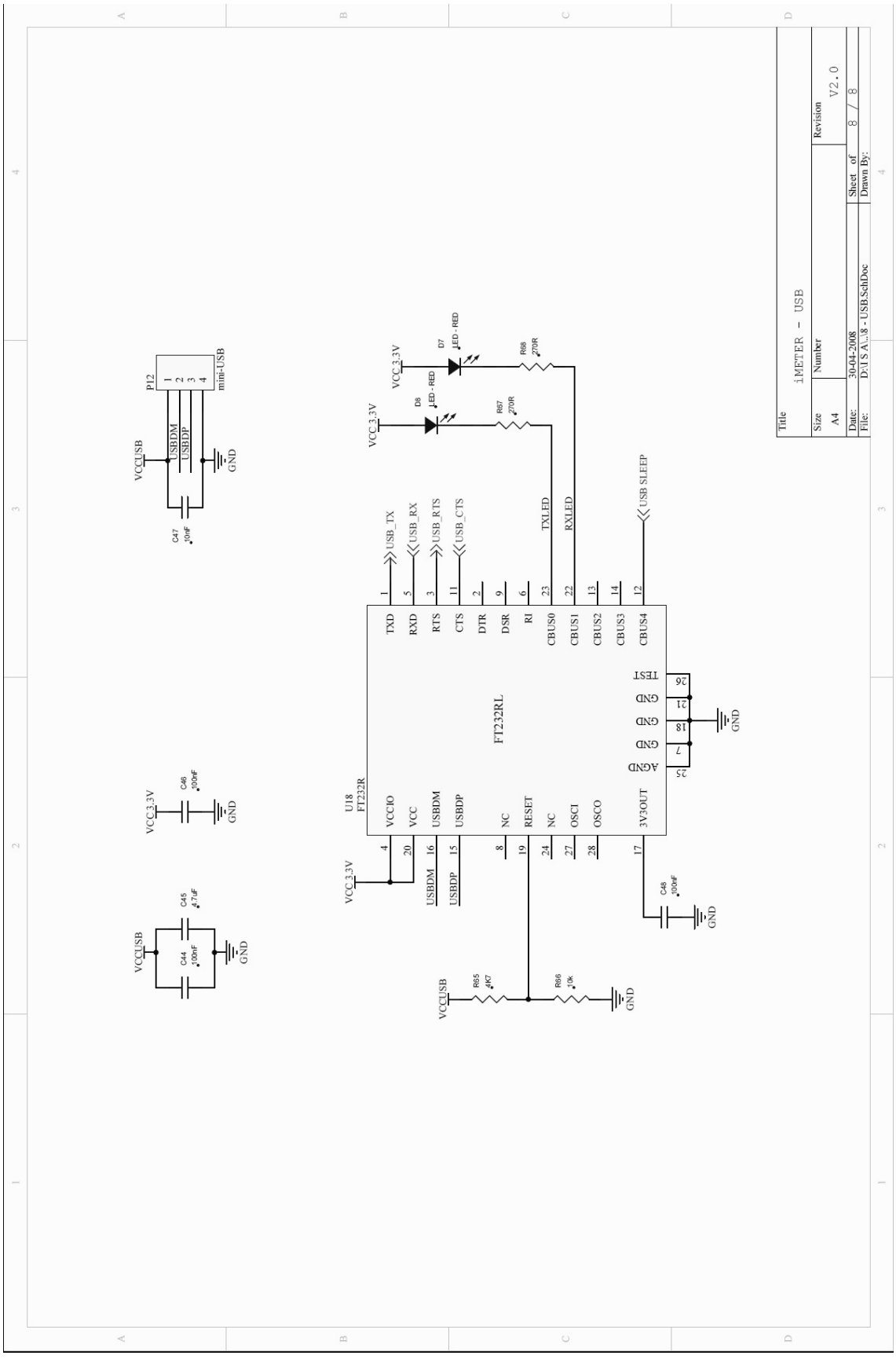
Size
A4
Number
Revision
V2.0
Date: 30-04-2008
Sheet of 6 / 8
File: D:\ISA\...\6 - GSM.SchDoc
Drawn By:
Modem TELIT
Modem WaveCom
SIM_Holder
VCC 3.3V
VCC 2.8V
VCC 2.75V
VCC_4V
VCC GSM
GND
TELIT OE
GSM OE
STATUS GSM
PWR ON
STAT LED
ON/OFF
LED GSM
SIM DATA
SIM CLK
SIM RST
SIM VCC
WCOM_RTS
WCOM_CTS
WCOM_RX
WCOM_TX
GSM IGNITION
TELITRXD
TELITTXD
TELITCTS
TELITRTS
CTRL POWER GSM
R28 220R
R51 10K
R52 2K2
R53 100k
R54 10k
R55 2K2
R56 100k
R57 10k
R58 2K2
R59 100k
R60 100k
R49 100K
C38 1uF
C39 1uF
C40 10nF
C41 .10uF
C42 6800uF
D9 LED0
Q3 BZ315BFS
Q4 FDV301
Q5 BC817 SOT 23
Q6 BC817 SOT 23
Q7 BC817 SOT 23
Q8 FDV301
U16 LP5951 SC-70
U17 SIM Holder SMD
P7
P8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Size
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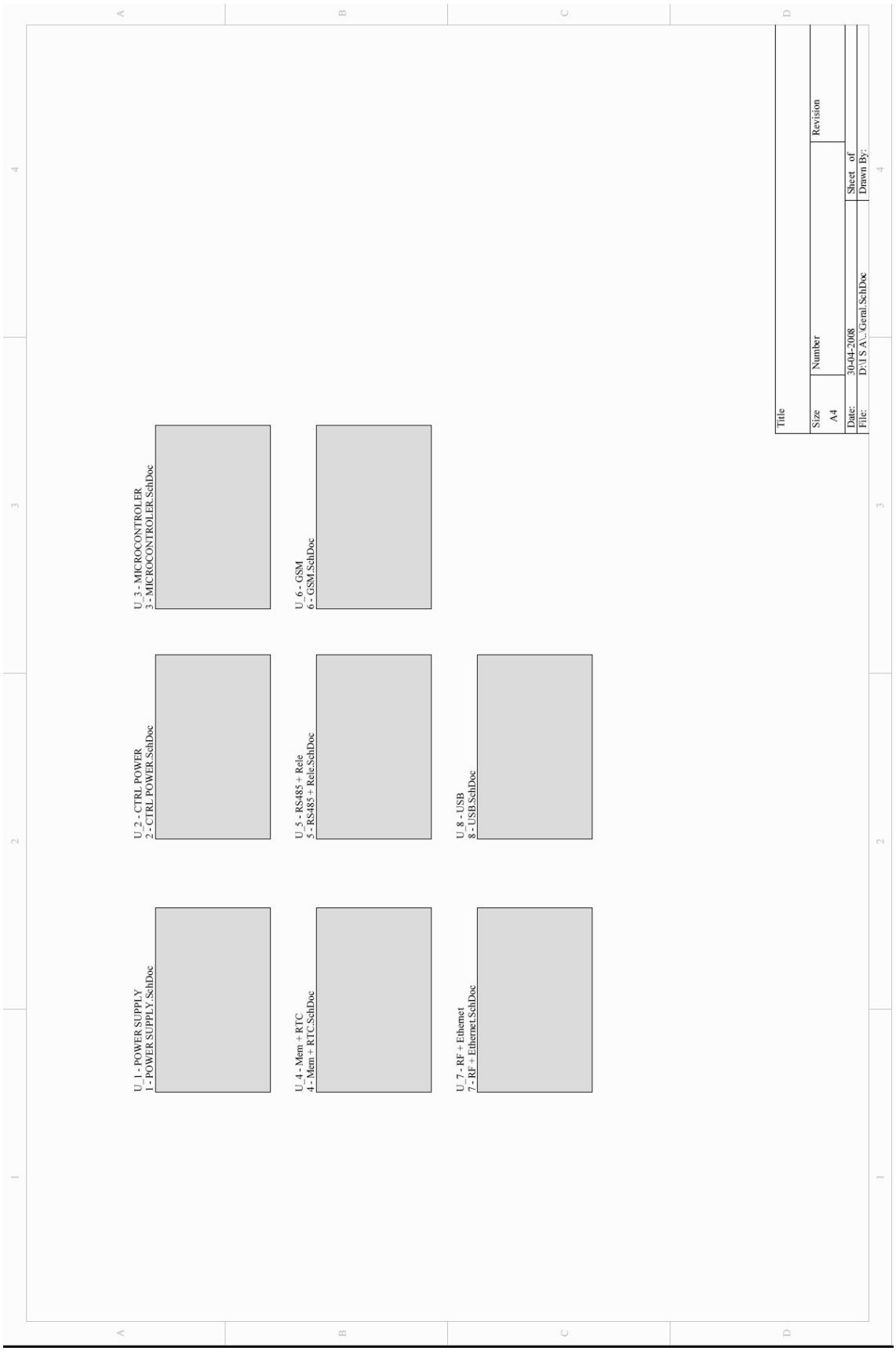
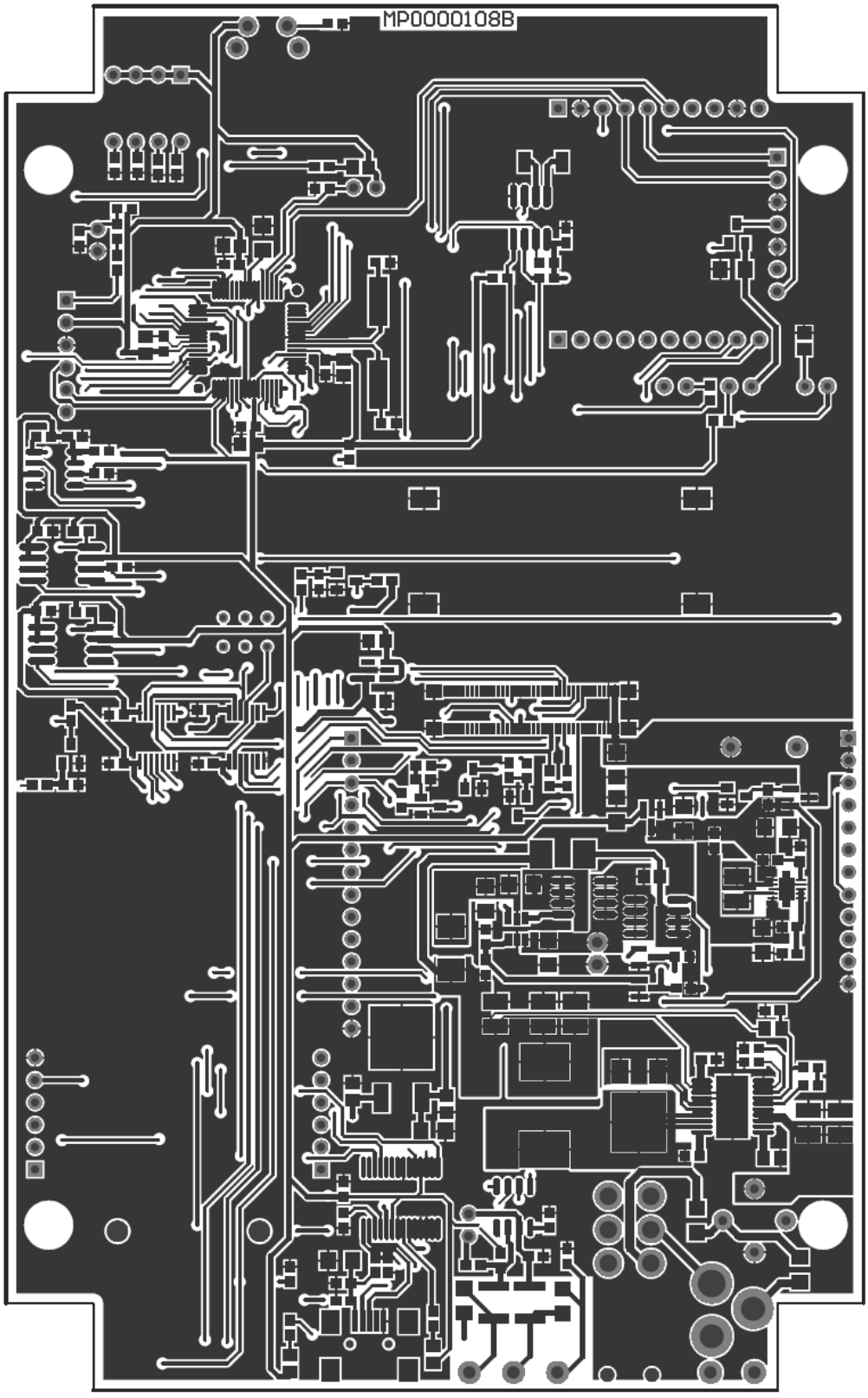
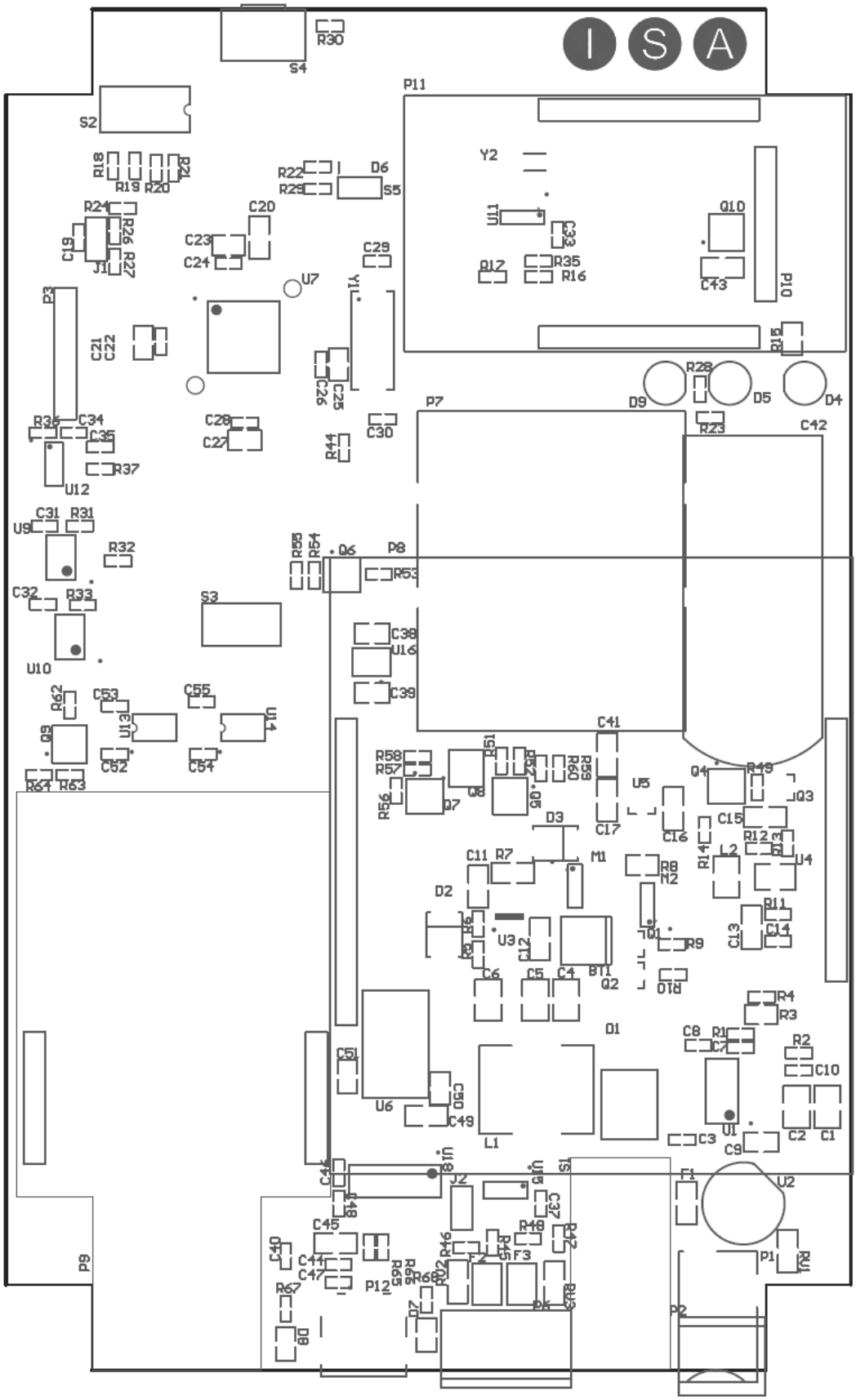
Number
Revision
V2.0
Date: 30-04-2008
Sheet of 8 / 8
File: D:\I S A\...8 - USB.SchDoc
Drawn By:
P12
mini-USB
USBDM
USBDP
VCCUSB
GND
C47
10nF
U18
FT232R
FT232RL
VCC3.3V
VCCIO
VCC
USBDM
USBDP
NC
RESET
OSCI
OSCO
3V3OUT
AGND
TEST
TXD
RXD
RTS
CTS
DTR
DSR
RI
CBUS0
CBUS1
CBUS2
CBUS3
CBUS4
USB_TX
USB_RX
USB_RTS
USB_CTS
TXLED
RXLED
USB SLEEP
D7
LED-RED
D8
LED-RED
R68
270R
R67
270R
R65
4K7
R66
10k
C48
100nF
C45
4.7uF
C44
100nF
C46
100nF

PCB's of the *Leonardo* Platform




MP0000108B


ISA
S4
R30
P11
S2
R18
R19
R20
R21
R22
R29
D6
S5
Y2
U11
C33
Q10
C43
P10
R24
R26
R27
C19
J1
C20
C23
C24
C29
U7
V1
R17
R35
R16
P3
C21
C22
C25
C26
R15
R28
D9
D5
D4
P7
R23
C42
R35
C34
C35
R37
U12
C28
C27
C30
R44
U9
C31
R31
R32
R55
R54
Q6
P8
R53
C32
R33
S3
C38
U16
C39
U10
R62
C53
C55
U13
U14
Q8
C52
C54
R64
R63
C41
R58
R57
R56
Q7
Q8
R52
R51
R50
R49
Q4
Q3
C17
C16
C15
U5
D3
C11
R7
M1
R8
R12
R13
L2
R14
U4
D2
U3
C12
B11
Q2
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R11
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R4
R3
D1
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C44
C47
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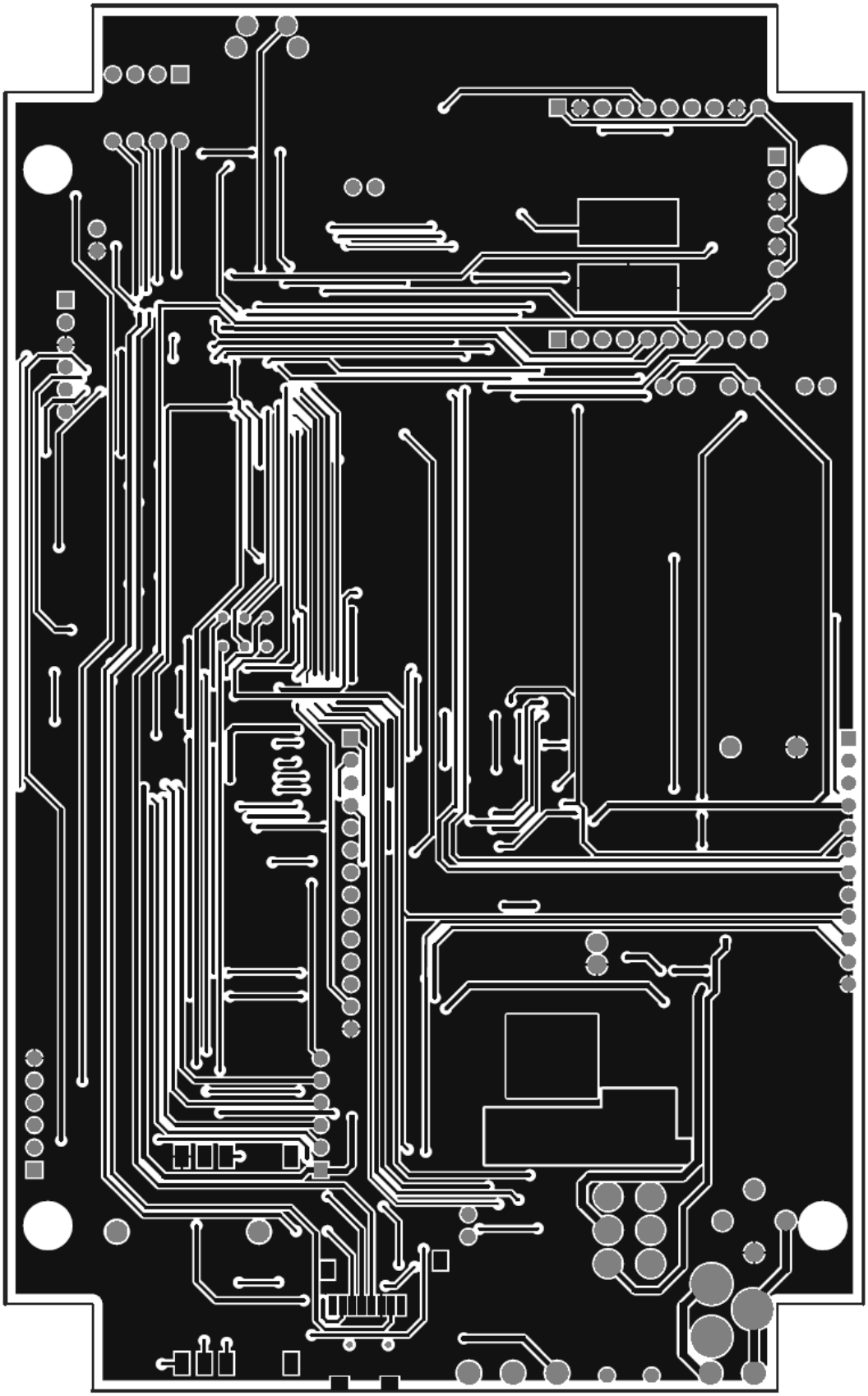
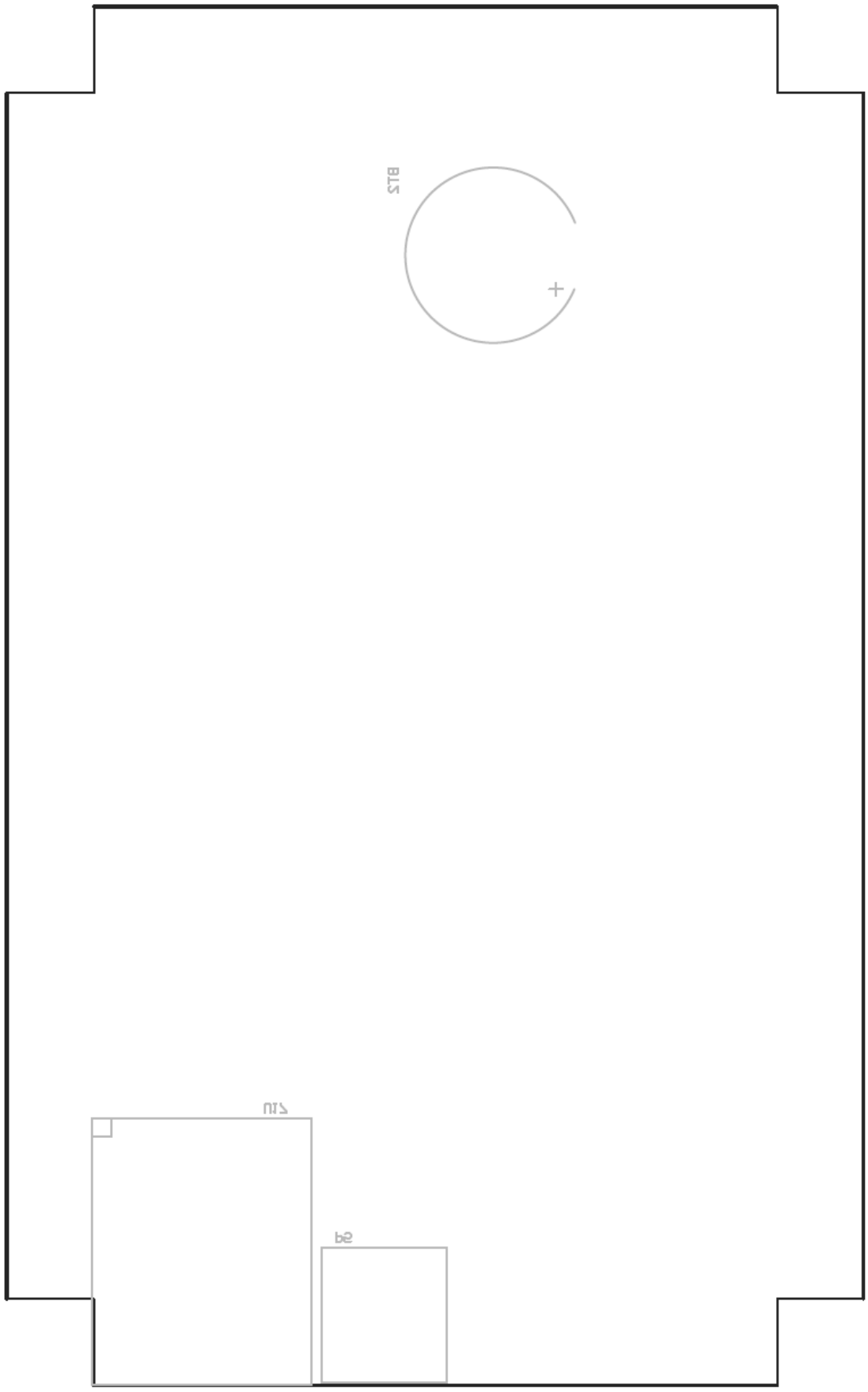
F3
P1
R41
P9
R65
R66
R62
P12
R67
D8
D7
R43
P2



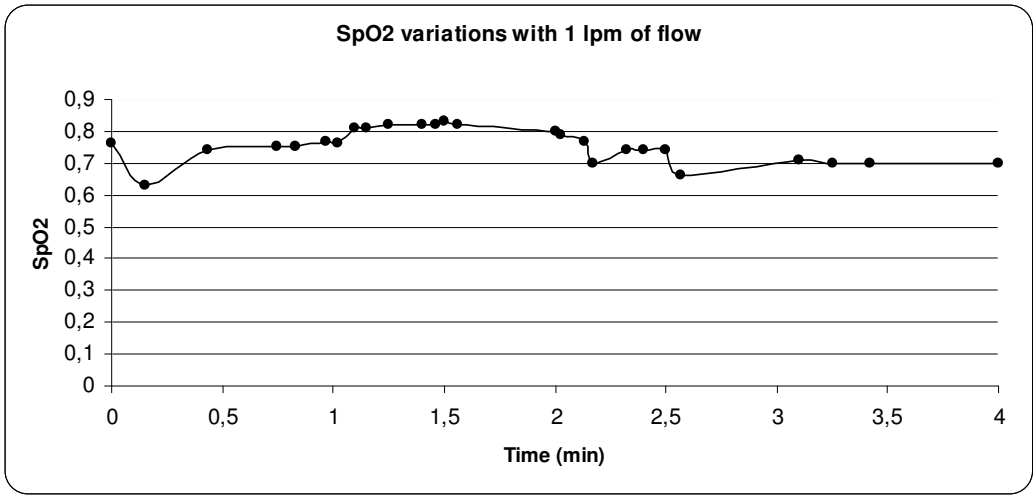

BT2
U12
P2

Patients' data collected at the Urgency Service

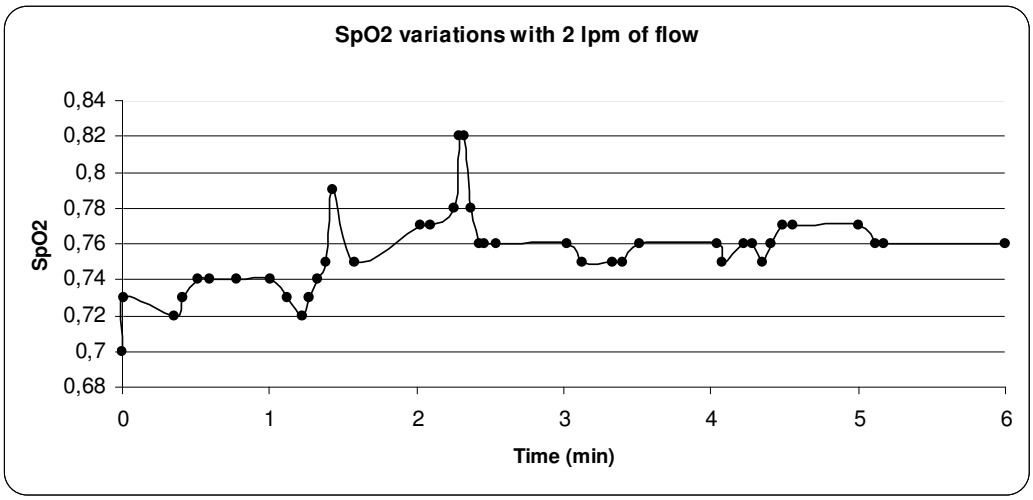
- **Patient with chronic lung edema (Male, 85 years)**

1 lpm		2 lpm		3 lpm		4 lpm		5 lpm		6 lpm	
Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2
0	0.76	0	0.70	0	0.76	0	0.80	0	0.84	0	0.85
9/60	0.63	1/60	0.73	2/60	0.76	1/60	0.80	1/60	0.84	1/60	0.85
26/60	0.74	21/60	0.72	12/60	0.76	13/60	0.80	17/60	0.83	31/60	0.86
45/60	0.75	25/60	0.73	32/60	0.76	22/60	0.81	28/60	0.84	38/60	0.86
50/60	0.75	31/60	0.74	40/60	0.77	28/60	0.81	1.05	0.84	44/60	0.86
58/60	0.77	36/60	0.74	49/60	0.77	38/60	0.80	1.19	0.84	1.07	0.86
1.02	0.76	47/60	0.74	51/60	0.78	43/60	0.79	1.30	0.84	1.13	0.85
1.10	0.81	1.01	0.74	1.12	0.78	47/60	0.80	1.43	0.83	1.19	0.85
1.15	0.81	1.12	0.73	1.24	0.77	53/60	0.80	1.51	0.84	1.25	0.85
1.25	0.82	1.22	0.72	1.29	0.78	57/60	0.80	1.57	0.85	1.32	0.86
1.40	0.82	1.27	0.73	1.43	0.78	1.01	0.79	2	0.84	1.40	0.86
1.46	0.82	1.33	0.74	1.58	0.79	1.10	0.81	2.16	0.84	1.44	0.84
1.50	0.83	1.39	0.75	2.03	0.79	1.16	0.82	2.23	0.85	1.50	0.84
1.56	0.82	1.43	0.79	2.11	0.79	1.22	0.80	2.32	0.85	1.56	0.84
2	0.80	1.58	0.75	2.20	0.80	1.27	0.79	2.40	0.85	2	0.84
2.03	0.79	2.03	0.77	2.25	0.80	1.35	0.80	2.58	0.84	2.20	0.85
2.13	0.77	2.09	0.77	2.39	0.79	1.42	0.80	3.05	0.85	2.25	0.85
2.17	0.70	2.25	0.78	2.44	0.78	1.49	0.80	3.17	0.84	2.30	0.86
2.32	0.74	2.29	0.82	2.50	0.79	1.54	0.81	3.30	0.84	2.35	0.86
2.40	0.74	2.33	0.82	2.55	0.80	2.01	0.81	4	0.84	2.40	0.86
2.50	0.74	2.37	0.78	3.03	0.80	2.22	0.82	4.10	0.85	2.50	0.85
2.57	0.66	2.43	0.76	3.09	0.79	2.29	0.83	4.12	0.84	3	0.85
3.10	0.71	2.46	0.76	3.14	0.80	2.30	0.82	4.22	0.84	3.15	0.85
3.25	0.70	2.54	0.76	3.20	0.80	2.40	0.82	4.30	0.85	4	0.85
3.42	0.70	3.02	0.76	3.34	0.81	2.54	0.83	4.41	0.86	4.06	0.84
4	0.70	3.13	0.75	3.43	0.80	3	0.83	4.47	0.86	4.10	0.84
		3.33	0.75	3.55	0.81	3.14	0.84	4.54	0.85	4.31	0.84
		3.40	0.75	3.59	0.80	3.18	0.84	4.58	0.84	4.36	0.85
		3.52	0.76	4.08	0.80	3.21	0.85	5.02	0.85	4.45	0.87
		4.04	0.76	4.20	0.80	3.26	0.84	5.10	0.84	4.50	0.87
		4.08	0.75	4.37	0.80	3.30	0.83	5.21	0.83	4.53	0.88
		4.23	0.76	4.49	0.80	3.38	0.84	5.29	0.85	4.58	0.88
		4.28	0.76	5	0.80	3.49	0.83	5.34	0.85	5.03	0.86
		4.35	0.75	5.11	0.81	3.54	0.84	5.38	0.85	5.06	0.87
		4.41	0.76	5.25	0.80	3.58	0.84	5.43	0.84	5.10	0.86
		4.49	0.77	5.30	0.80	4.04	0.84	5.46	0.84	5.46	0.86
		4.56	0.77	5.40	0.81	4.16	0.83	5.51	0.85	5.56	0.85
		5	0.77	5.47	0.80	4.30	0.83	6.02	0.85	6.15	0.86
		5.12	0.76	5.58	0.80	4.50	0.84	6.08	0.85	6.20	0.85
		5.20	0.76	6	0.80	5	0.84	6.20	0.85	6.30	0.85
		6	0.76					6.30	0.86	6.40	0.84
								6.42	0.85	6.44	0.81
								7	0.85	6.46	0.80
										6.57	0.82
										6.7	0.82
										7.06	0.82
										7.10	0.83
										7.25	0.83
										7.31	0.85
										7.41	0.86
										7.50	0.86
										8	0.86

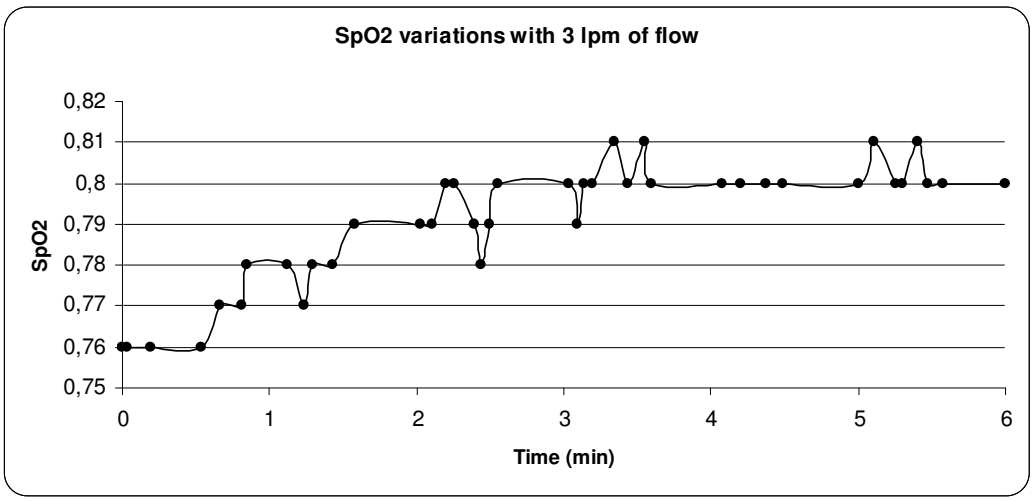
Table 10: Data from a patient with chronic lung edema, the SpO_2 variations with oxygen flows from 1 to 6 lpm.



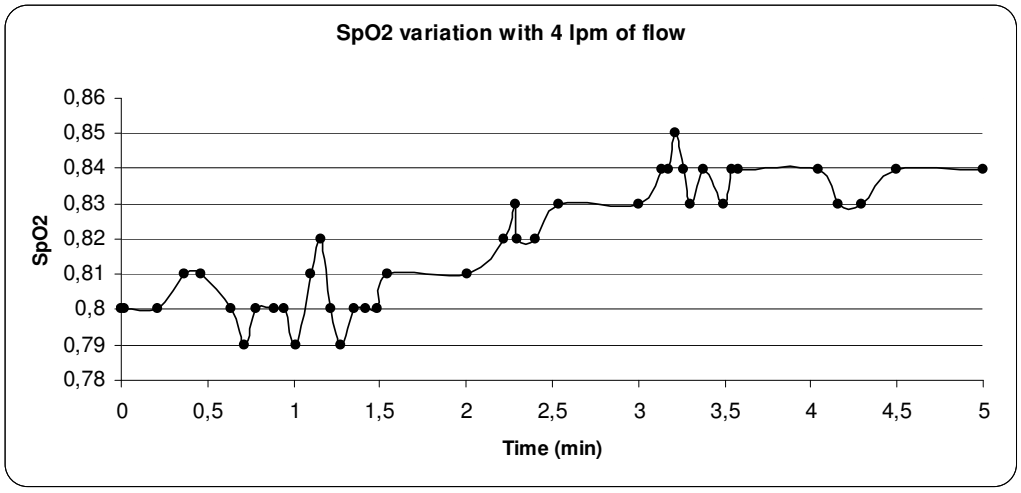
Graphic 4: SpO_2 variations from patient 1, with 1 lpm of oxygen flow.



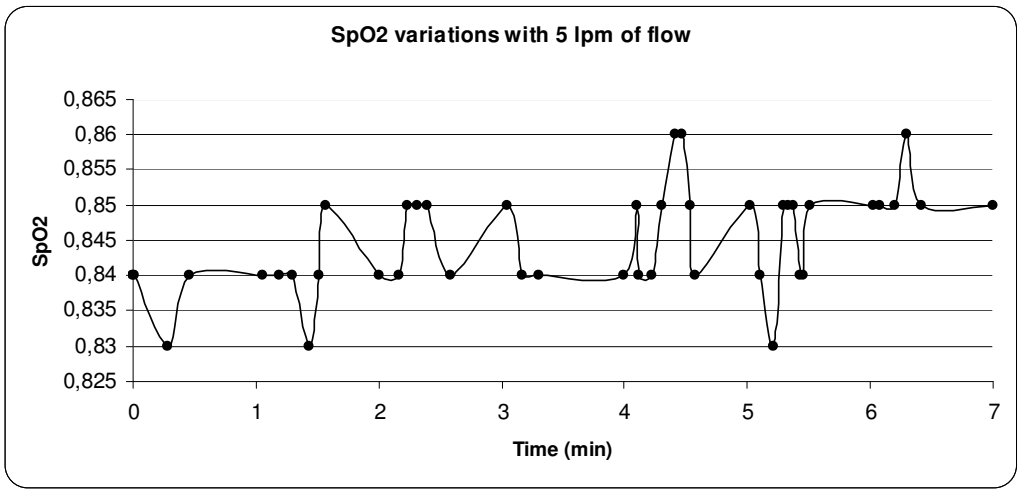
Graphic 5: SpO_2 variations from patient 1, with 2 lpm of oxygen flow.



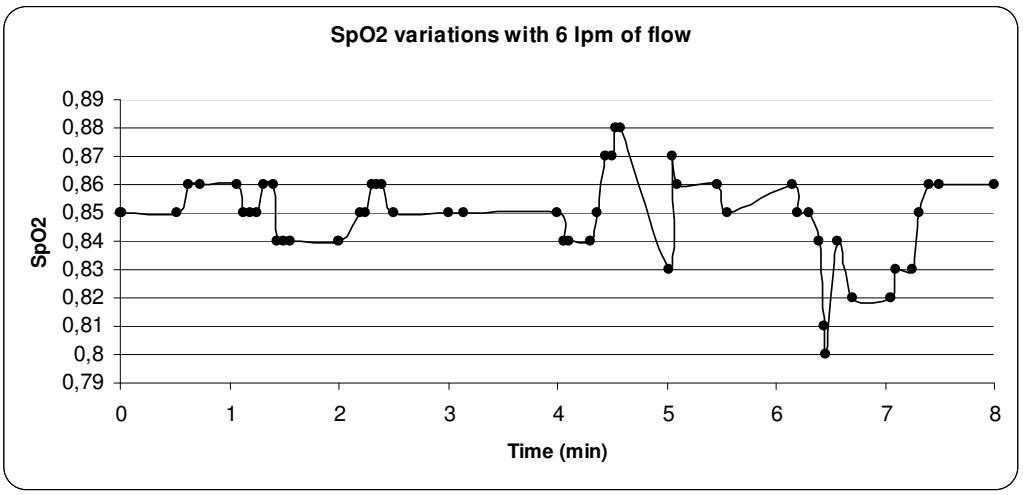
Graphic 6: SpO_2 variations from patient 1, with 3 lpm of oxygen flow.



Graphic 7: SpO_2 variations from patient 1, with 4 lpm of oxygen flow.



Graphic 8: SpO_2 variations from patient 1, with 5 lpm of oxygen flow.



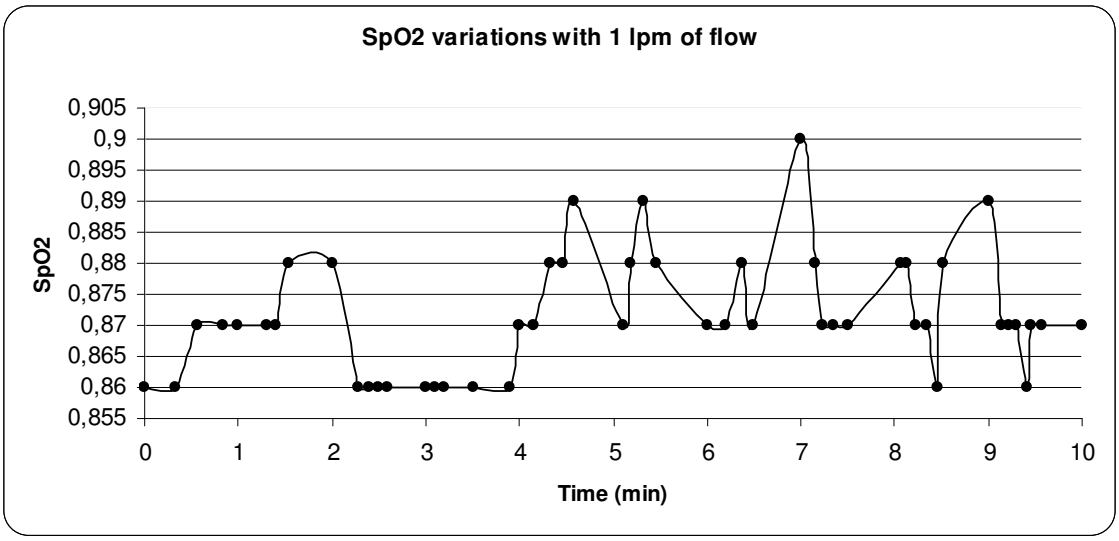
Graphic 9: SpO_2 variations from patient 1, with 6 lpm of oxygen flow.

- **Patient with diabetes and obesity (Male, 79 years)**

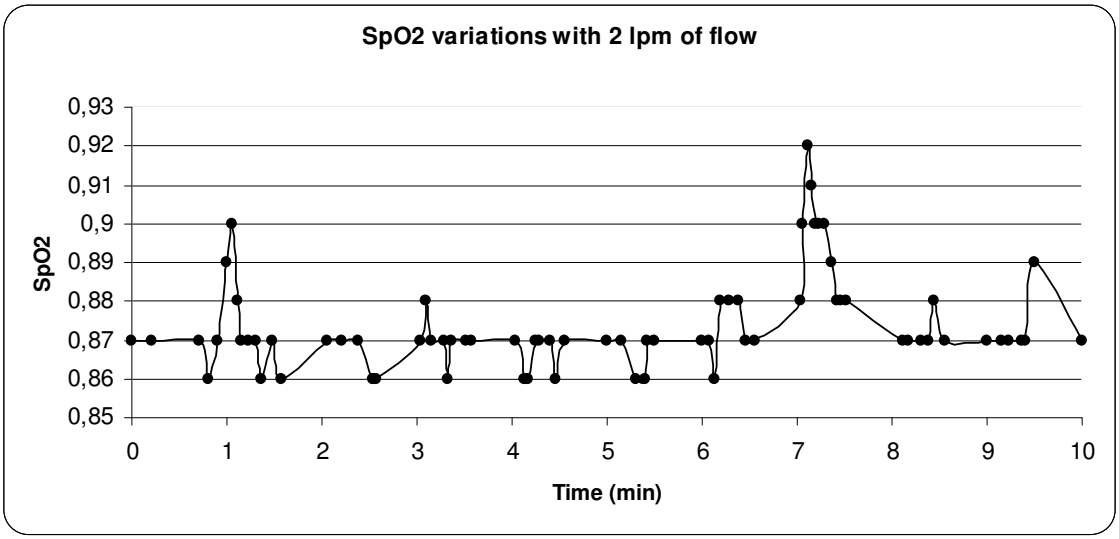
1 lpm		2 lpm		3 lpm		4 lpm	
Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2	Time (min)	SpO_2
0	0.86	0	0.87	0	0.87	0	0.88
20/60	0.86	13/60	0.87	7/60	0.88	11/60	0.88
34/60	0.87	43/60	0.87	12/60	0.88	24/60	0.89
50/60	0.87	48/60	0.86	26/60	0.88	32/60	0.89
1	0.87	54/60	0.87	36/60	0.87	46/60	0.89
1.3	0.87	1	0.89	41/60	0.90	1	0.89
1.4	0.87	1.05	0.90	44/60	0.91	1.18	0.89
1.54	0.88	1.11	0.88	54/60	0.90	1.25	0.89
2	0.88	1.16	0.87	58/60	0.89	1.30	0.89
2.28	0.86	1.24	0.87	1.07	0.88	1.46	0.89
2.4	0.86	1.30	0.87	1.14	0.88	1.55	0.89
2.5	0.86	1.37	0.86	1.28	0.88	2	0.89
2.6	0.86	1.48	0.87	1.37	0.88	2.05	0.88
3	0.86	1.57	0.86	1.45	0.88	2.15	0.88
3.1	0.86	2.05	0.87	1.51	0.88	2.26	0.88
3.2	0.86	2.22	0.87	2	0.88	2.40	0.88
3.5	0.86	2.39	0.87	2.15	0.88	3	0.88
3.9	0.86	2.54	0.86	2.23	0.88	3.19	0.88
4	0.87	2.58	0.86	2.38	0.88	3.24	0.89
4.15	0.87	3.04	0.87	2.41	0.89	3.36	0.89
4.32	0.88	3.10	0.88	2.48	0.89	3.54	0.89
4.46	0.88	3.16	0.87	2.54	0.89	3.57	0.90
4.54	0.89	3.28	0.87	2.57	0.90	4.04	0.91
5.11	0.87	3.33	0.86	3.03	0.90	4.09	0.90
5.19	0.88	3.37	0.87	3.07	0.89	4.16	0.89
5.32	0.89	3.51	0.87	3.13	0.89	4.25	0.88
5.46	0.88	3.57	0.87	3.21	0.89	4.30	0.89
6	0.87	4.04	0.87	3.25	0.88	4.48	0.89
6.20	0.87	4.13	0.86	3.31	0.88	5	0.89
6.37	0.88	4.18	0.86	3.39	0.88	5.10	0.89
6.50	0.87	4.25	0.87	3.44	0.89	5.20	0.88
7	0.90	4.29	0.87	3.51	0.89	5.27	0.88
7.15	0.88	4.40	0.87	3.56	0.89	5.38	0.88
7.23	0.87	4.47	0.86	4.02	0.88	5.50	0.88
7.35	0.87	4.55	0.87	4.13	0.88	6	0.88
7.50	0.87	5	0.87	4.20	0.89	6.14	0.89
8.07	0.88	5.15	0.87	4.25	0.89	6.28	0.89
8.13	0.88	5.30	0.86	4.30	0.90	6.40	0.89
8.23	0.87	5.40	0.86	4.35	0.90	6.53	0.89
8.35	0.87	5.42	0.87	4.41	0.91	7.03	0.89
8.46	0.86	5.50	0.87	4.46	0.92	7.07	0.88
8.52	0.88	6	0.87	4.51	0.92	7.17	0.88
9	0.89	6.08	0.87	4.53	0.91	7.29	0.88
9.14	0.87	6.14	0.86	4.57	0.91	7.35	0.88
9.23	0.87	6.20	0.88	5.02	0.90	7.56	0.88
9.30	0.87	6.28	0.89	5.07	0.90	8	0.88
9.42	0.86	6.38	0.88	5.10	0.89	8.21	0.88
9.46	0.87	6.47	0.87	5.17	0.89	8.27	0.89
9.57	0.87	6.56	0.87	5.29	0.89	8.31	0.87
10	0.87	7.03	0.88	5.38	0.89	8.41	0.87
		7.06	0.90	5.50	0.89	8.49	0.86
		7.11	0.92	5.57	0.88	8.59	0.86
		7.15	0.91	6	0.88	9	0.85
		7.19	0.90	6.07	0.89	9.11	0.85
		7.24	0.90	6.14	0.89	9.17	0.87
		7.29	0.90	6.25	0.89	9.22	0.90
		7.36	0.89	6.52	0.89	9.27	0.91
		7.42	0.88	7	0.89	9.34	0.91

2 lpm (cont.)		3 lpm (cont.)		4 lpm (cont.)	
7.47	0.88	7.03	0.90	9.44	0.90
7.51	0.88	7.18	0.90	9.54	0.90
8.11	0.87	7.25	0.89	10	0.90
8.17	0.87	7.30	0.90		
8.30	0.87	7.37	0.90		
8.39	0.87	7.40	0.90		
8.45	0.88	7.46	0.89		
8.55	0.87	7.50	0.89		
9	0.87	8.04	0.90		
9.16	0.87	8.10	0.91		
9.23	0.87	8.15	0.94		
9.37	0.87	8.24	0.95		
9.41	0.87	8.28	0.94		
9.50	0.89	8.33	0.93		
10	0.88	8.36	0.92		
		8.40	0.92		
		8.46	0.92		
		8.48	0.90		
		9	0.89		
		9.07	0.89		
		9.15	0.88		
		9.35	0.88		
		9.42	0.88		
		9.47	0.89		
		9.54	0.88		
		10	0.88		

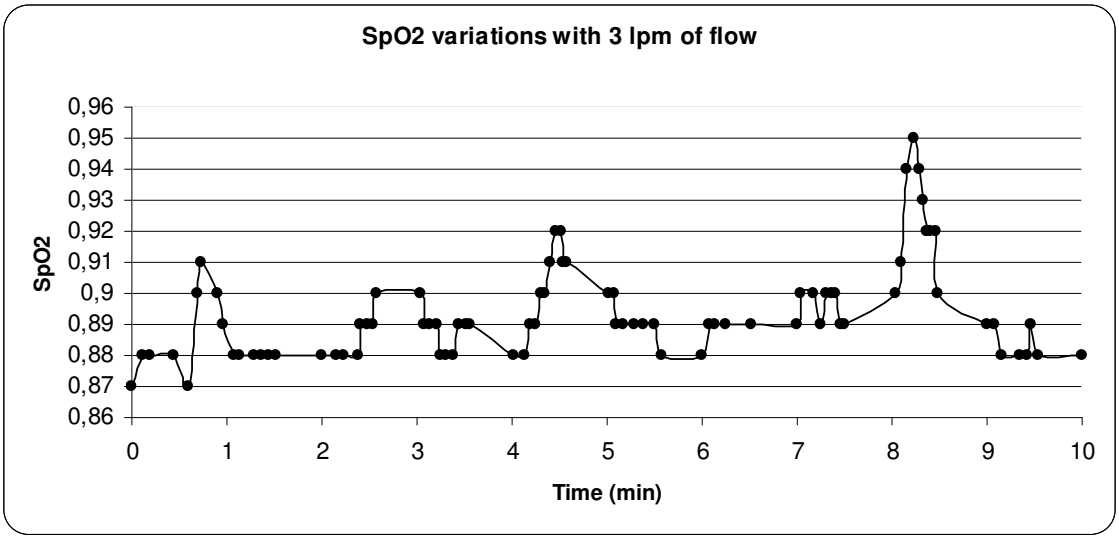
Table 11: Data from a patient with diabetes and obesity, the SpO_2 variations to oxygen flows from 1 to 4 lpm.



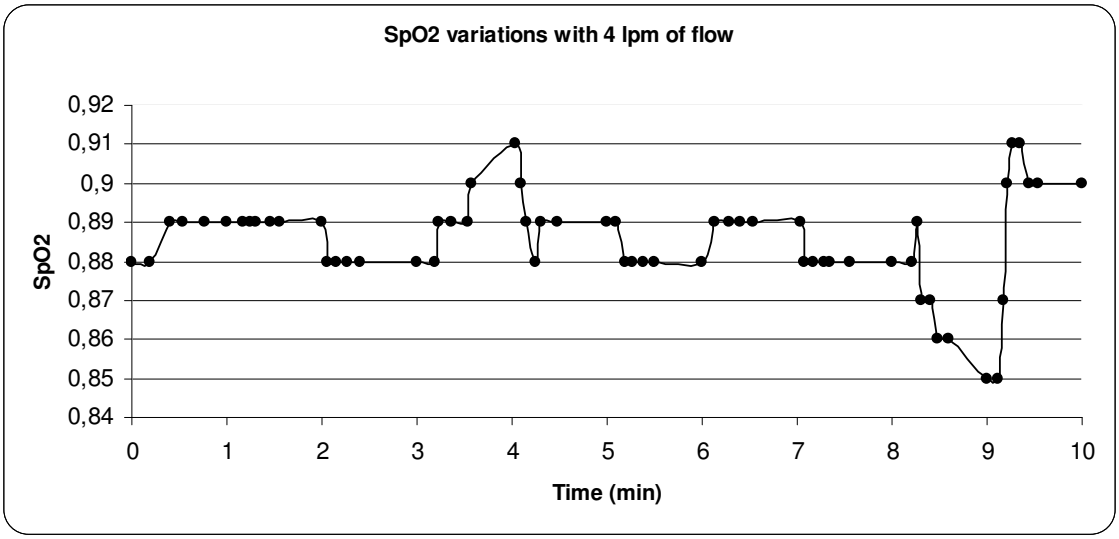
Graphic 10: SpO_2 variations from patient 2, with 1 lpm of oxygen flow.



Graphic 11: SpO_2 variations from patient 2, with 2 lpm of oxygen flow.



Graphic 12: SpO_2 variations from patient 2, with 3 lpm of oxygen flow.



Graphic 13: SpO_2 variations from patient 2, with 4 lpm of oxygen flow.

Driver Board Technical Specifications

Kelly Pneumatics
Pressure & Flow Control Solutions

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Proportional Valve Driver Board

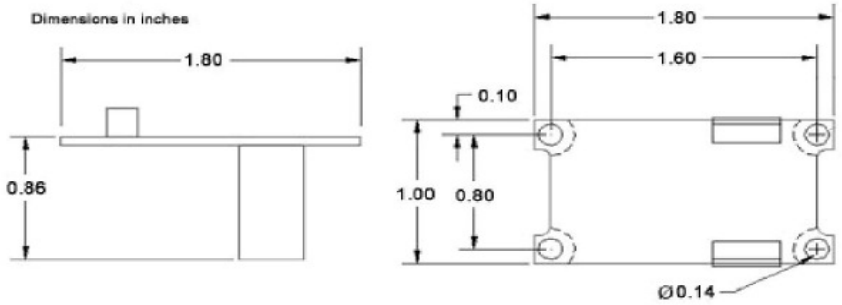
Build Part

- Proportional Voltage Output
- High Resolution Current Output
- Usable in Variety of Electronic Assemblies

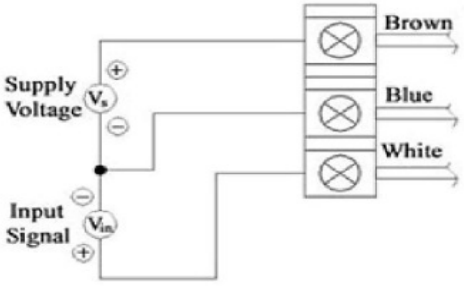
Kelly Pneumatics, Inc. offers a Driver Board for its Proportional Valve product line. This driver board converts an input signal proportionally to an output current, driving the attached valve appropriately with excellent resolution. Can also be applied to other electronic applications in need of precise current control.

Voltage Required	12 or 24 VDC +/- 10%
Current Output	< 3 watts (Mini-Proportional Valve) < 6 watts (Midsize Proportional Valve) varies for other products
Input Signals	0-5 VDC, 0-10 VDC, 4-20 milliamps
Operating Temp	32-150 °F or 0-65 °C

Dimensions



Wiring



Terminal Number	1	2	3
Lead Wire Color	Brown	White	Blue
Wiring	Power Supply	Input Signal	Common (Ground)

Miniature Proportional Valve Technical Specifications

Kelly Pneumatics
Pressure & Flow Control Solutions

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Mini Proportional Valve

Build Part

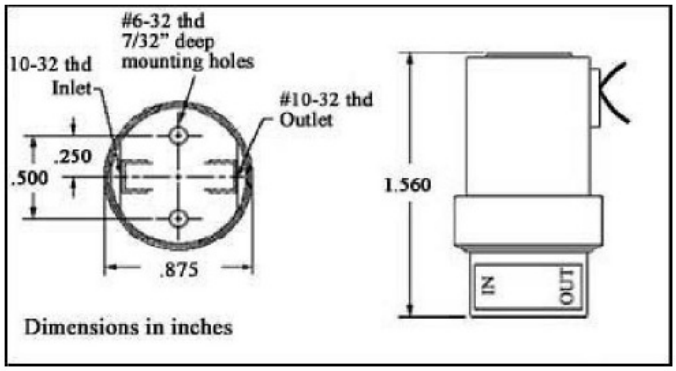
- Precise Volume Outlet Flow Capability
- Easily Cleaned and Maintained
- Low Hysteresis Performance
- Controls Air, Water, and Other Mediums

Kelly Pneumatics, Inc. offers a low volume flow Miniature Proportional Valve utilizing award winning design. This non-spool valve architecture employing one moving part ensures virtually frictionless performance and longevity tested at over 100 million cycles. This simplicity in design also assures relative ease when implementing maintenance or cleaning. We also offer alternate versions that are resistant to both high and low temperatures, therefore performing consistently in many different environments. Available in both 2-way and 3-way versions, this valve can efficiently process Water, Air, Oil and Hydrogen.

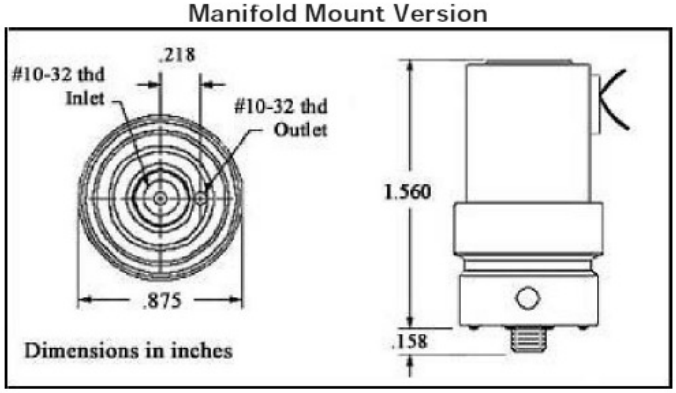
Valve Type	2-Way or 3-way, Proportional
Electrical Connection	Wire Leads or Terminal Spades
Working Pressure	25, 50, and 100 psig
Flow Range	0-500 milliliters of water per minute or 0-20 slpm of air flow (lower flow values available)
Orifice Size	.009, .013, .025, .040, or .060 inches
Voltage Range	0-5, 0-10, and 0-20 VDC @ 2.3 watts max
Response Time	3 - 5 ms
Port Types	#10-32 Inline (Base Mount), #10-32 Manifold Mount, or Pipe Threads (See Dimensions Below)
Seal Material	Buna-N, Viton, or EPDM
Operating Temp.	0-150 ^{0}F or 0-65 ^{0}C
Maximum Hysteresis	10 % of full current
Ports	See Dimensions Below

Dimensions and Port Locations

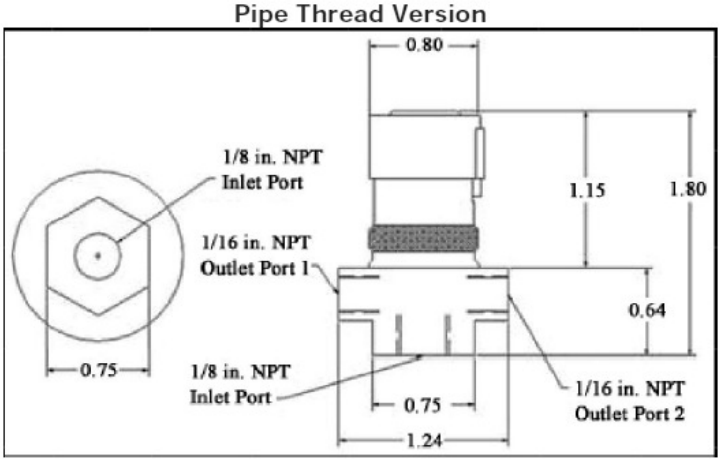
Base Mount Version



Manifold Mount Version

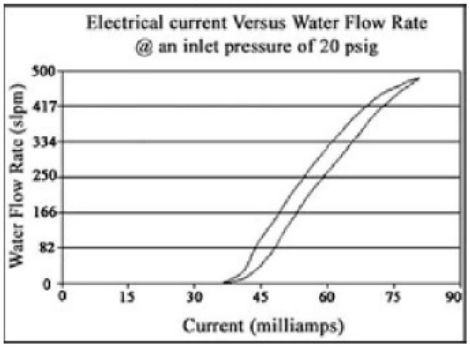


Pipe Thread Version

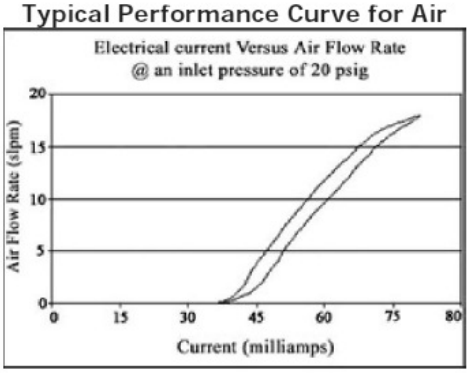


Performance Characteristics

Typical Performance Curve for Water



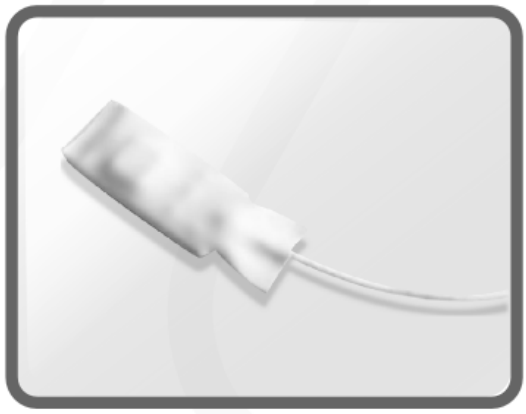
Typical Performance Curve for Air



XYZ Plux Accelerometer Technical Specifications

xyzPlux

triaxial accelerometer



technical specifications

- **±3G** measurement range
- **3** meassurement axis
- **MEMS®** technology
- **10000G** shock resistance

description

Accelerometer

Detect sudden motion changes in micro electronic mechanical systems (MEMS) in three axys.

Applications:

.: vibration analysis
.: motion paterns
.: walking behaviour
.: fall detection
.: tilt

OxiLink™ Pulse Oximeter Technical Specifications

BCI® OxiLink™ Specifications

SPECIFICATIONS	OxiLink™
SpO_2	
Range	0-99% (1% increments)
Accuracy (Specifications may vary depending on sensor used)	Adult: ±2% at 70-99% SpO_2 <70% is undefined Neonate: ±3% at 70-99% SpO_2 <70% is undefined
Averaging	8 pulse beat average and instantaneous
Pulse Rate	
Range	30-254 BPM (1 BPM increments)
Accuracy	±2 BPM or ±2% (whichever is greater)
Averaging	8 second average
Signal Strength	0-8 indicates logarithmic strength of patient's pulse from 30-254 BPM
Bargraph	0-15 segments
Patient Isolation	1.5kV, provided by device
Power Requirements	As low as 25 mW
Dimensions	3.25" W (8.3 cm) 1.08" D (2.7 cm) 0.57" H (1.5 cm)
Flags	Pulse beep No finger in sensor Sensor unplugged Searching for pulse Searching too long
Communication/Power Options	
Serial	RS232 USB